Robotic Functional Gait Rehabilitation with Tethered Pelvic Assist Device

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Abstract

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The primary goal of human locomotion is to stably translate the center of mass (CoM) over the ground with minimum expenditure of energy. Pelvic movement is crucial for walking because the human CoM is located close to the pelvic center. Because of this anatomical feature, pelvic motion directly contributes to the metabolic expenditure, as well as in the balance to keep the center of mass between the legs. An abnormal pelvic motion during the gait not only causes overexertion, but also adversely affects the motion of the trunk and lower limbs. In order to study different interventions, recently a cable-actuated robotic system called Tethered Pelvic Assist Device (TPAD) was developed at ROAR laboratory at Columbia University. The cable-actuated system has a distinct advantage of applying three dimensional forces on the pelvis at discrete points in the gait cycle in contrast to rigid exoskeletons that restrict natural pelvic motion and add extra inertia from the rigid linkages. However, in order to effectively use TPAD for rehabilitation purposes, we still need to have a better understanding of how human gait is affected by different forces applied by TPAD on the pelvis. In the present dissertation, three different control methodologies for TPAD are discussed by performing human experiments with healthy subjects and patients with gait deficits. Moreover, the corresponding changes in the biomechanics during TPAD training are studied to understand how TPAD mechanistically influences the quality of the human gait.

In Chapter 2, an 'assist-as-needed' controller is implemented to guide and correct the pelvic motion in three dimensions. Here, TPAD applies the correction force based on the deviation of the current position of the pelvic center from a pre-defined target trajectory.

This force acts on the pelvic center to guide it towards the target trajectory. A subject in the device experiences a force field, where the magnitude becomes larger when the subject deviates further away from the target trajectory. This control strategy is tested by performing the experiments on healthy subjects with different target pelvic trajectories.

Chapter 3 describes a robotic resistive training study using a continuous force on the pelvis to strengthen the weak limbs so that subjects can improve their walking. This study is designed to improve the abnormal gait of children with Cerebral Palsy (CP) who have a crouch gait. Crouch gait is caused by a combination of weak extensor muscles that do not produce adequate muscle forces to keep the posture upright, coupled with contraction of muscles that limit the joint range of motion. Among the extensor muscles, the soleus muscle acts as the major weight-bearing muscle to prevent the knees from collapsing forward during the middle of the stance phase when the foot is on the ground. Electromyography, kinematics, and clinical measurements of the patients with crouch gait show significant improvements in the gait quality after the resistive TPAD training performed over five weeks.

Both Chapters 2 & 3 present interventions that are bilaterally applied on both legs. Chapter 4 introduces a training strategy that can be used for patients who have impairments in only one leg which results in manifests as asymmetric weight-bearing while walking. This training method is designed to improve the asymmetric weight bearing of the hemiparetic patients who overly rely on the stronger leg. The feasibility of this training method is tested by experiments with healthy subjects, where the controller creates an asymmetric force field to bring asymmetry in weight bearing during walking.

In summary, the present dissertation is devoted to developing new training methods that utilize TPAD for rehabilitation purposes and characterize the responses of different force interventions by investigating the resulting biomechanics. We believe that these methodologies with TPAD can be used to improve abnormal gait patterns that are often observed in cerebral palsy or stroke patients.

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Dedication

I would like to dedicate this thesis to God

and my beloved husband, family, and friends

who showed endless love and support during my Ph. D. program.

Chapter 1

Background

1.1 Introduction: gait rehabilitation and role of the pelvis

The goal of human locomotion is to displace the body's center of mass (CoM) along the direction of motion [60]. For humans, the CoM motion is optimized to reduce the metabolic cost while maintaining stability on the legs [66, 69, 82]. In order to achieve this goal in human locomotion, the CoM moves in a complex trajectory. The CoM lies approximately at the center of the pelvis, in front of the second sacral vertebra, and pelvic center motion is often approximated as the CoM movement [125, 114, 122]. As the motion of the CoM fundamentally contributes to the metabolic cost, in order to accelerate/decelerate the body and raise it against the gravity [118], the trajectory of pelvic movement is related to the energy efficiency of the gait.

The pelvic center can be defined by the mid point of three anatomical landmarks on the pelvis, right/left iliac crest and sacrum, which are observed by a motion capture system. It forms the shape of a butterfly wings, once projected in two dimensional plane consisting of vertical (y axis) and medial-lateral (x axis) axes, as seen by the green curve in Fig.1.1 [49]. This curve can be separated into two sub-plots along the gait cycle for each vertical and medial-lateral displacement. In a gait cycle, vertical motion of the pelvic center can be

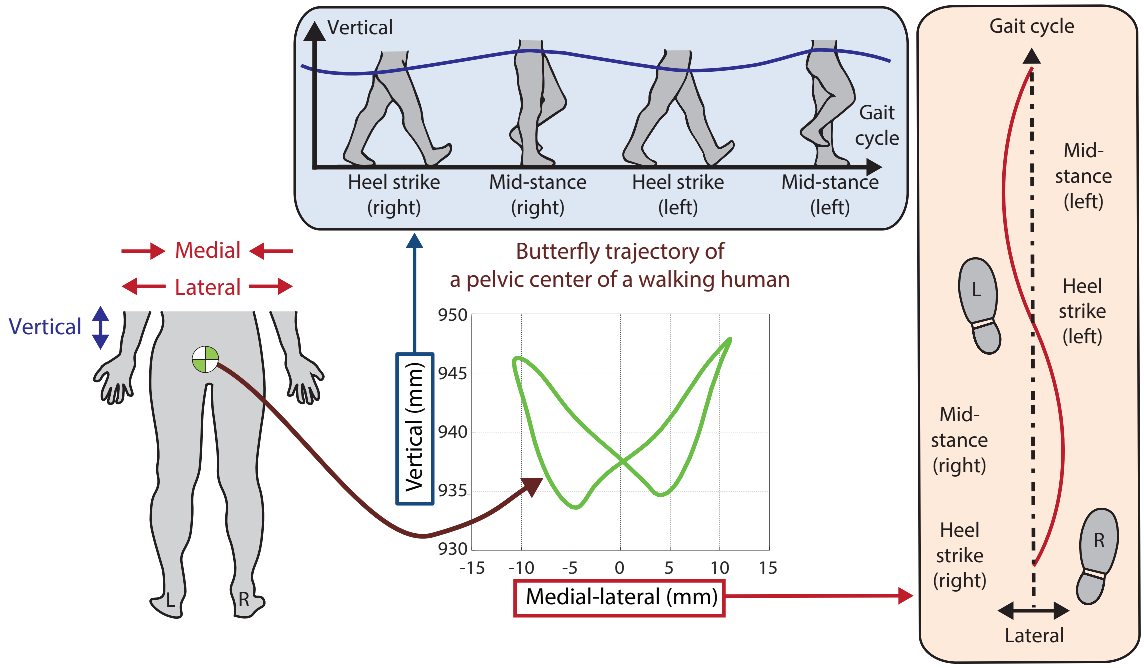


Fig. 1.1 Pelvic center displacement in the shape of butterfly wings that is plotted by green curve in two dimensional space of vertical and medial-lateral axes during a gait cycle [49, 71, 78]. This graph can be divided into two sub-plots: vertical (blue box) and medial-lateral (red box) displacement of the pelvic center over the gait cycle.

expressed as a sinusoid wave with two peaks over a gait cycle, as seen on the top of Fig. 1.1 [71]. During the initial stance phase after the heel strike, legs raise the COM until the mid stance phase of the gait. Then, the COM moves downward and forward, converting potential energy to kinetic energy [13]. As one gait cycle consists of two stance phases, one for each right and left foot, the vertical pelvic trajectory presents two peaks when each leg is straight.

The medial-lateral motion of the pelvis enables the bipedal locomotion by shifting the weight to one side of the leg [78]. The red curve in Fig. 1.1 presents the lateral sway of the pelvic center during the gait cycle, which shifts the weight on the right leg and then to the left [64]. The lateral motion can be approximated as one period of sinusoidal wave. Each half of this sinusoidal wave starts at each leg's heel strike and moves towards that leg. The lateral pelvic motion can be used as the indicator of lateral weight shift because the lateral motion of the pelvis is synchronized with the center of gravity of the human body [114]. Unlike sagittal motion, human actively controls the lateral pelvic motion to maintain stability, which is responsible for the majority of metabolic energy expenditure [30].

Because pelvic motion has various roles in the human gait as described so far, pelvic motion is often used by clinicians to detect the pathological gait pattern [94]. Children who are diagnosed with cerebral palsy have gait impairments. There are about 3.5 out of 1000 children in US who are diagnosed with cerebral palsy and some of these patients present a gait pattern called crouch gait. Couch gait is a movement disorder that presents excessive dorsi-flexion, hip flexion, and knee flexion which is caused by tight flexor muscles. These children are observed with lower vertical pelvic height due to overly flexed knee joints [103]. Besides the vertical motion, larger lateral body sway is often observed in these children because they have difficulty in controlling the peripheral muscle of the ankle or knee to clear the foot from the ground [43, 93, 34]. This excessive sway of the CoM leads to early fatigue and higher energy expenditure. This abnormal gait pattern is often observed in patients with stroke as well. However, they also have the asymmetric pattern in the lateral pelvic motion because they are impaired on one leg, also observed in stroke patients [28]. This asymmetry in lateral pelvic motion leads to overloading of the stronger limb and loss of muscle strength on the paretic limb [55, 84].

1.2 State-of-art of pelvic assist devices

Last few decades, researchers have developed devices for lower limb while recognizing the importance of the pelvic motion during the gait. They developed novel robotic mechanisms to support and control the pelvic motion. In 2005, Peshkin et al. [86] designed a robotic gait/balance training device with a mobile base that supports both the trunk and the pelvis. The trunk module part consists of two active DoFs for lateral/forward bending of the trunk, one passive DoF for axial rotation, and one passive vertical translation. However, for the pelvis, it has five passive degree of freedom while excluding the vertical translation. As

Fig. 1.2 Different robotic designs for pelvic support or assistance of (a) KineAssist developed by Peshkin et al. in 2005 [86], (b) Pelvic Assist Manipulator by Aoyagi et al. in 2007 [8], (c) Robotic Gait Rehabilitation trainer by Pietrusinski et al. in 2014 [87], (d) WalkTrainer by Allemand et al. in 2009 [6], (e) NaTUre-gaits by Luu et al. in 2014 [63], (f) body weight support system by Mun et al. in 2017 [75], (g) balance assessment robot (BAR) by Shirota et al. in 2017 [100], (h) the cable-driven robotic system by Hsu et al. in 2017 [42], and (i) AssistOn-Gait by Munawar et al. in 2016 [76].

an actively controlled pelvic device, Aoyagi et al. [8] developed a Pelvic Assist Manipulator (PAM) in 2007 with pneumatic actuators to support full pelvic range of motion. However, this device used position control to move the pelvis in a pre-defined trajectory that has limitations to encourage active participation of patients during the training [8]. For facilitating interactive robot–patient control, Pietrusinski et al. in 2014 designed Robotic Gait Rehabilitation (RGR) trainer [87] that was controlled based on the 'assist-as-needed' concept that provided assistance based on their motion [16]. RGR trainer applied an assistive force on the pelvis to correct pelvic obliquity (rotation about the antero-posterior axis) and its feasibility was tested on healthy subjects to mimic hip-hiking of stroke patients.

Mobile type pelvic devices have also been developed by adding active pelvic modules as seen in Fig. 1.2(d-g). Mun et. al. [75] added a holonomic mobile base to the walker which controls the interactive forces between the patient and the device. For facilitating a human-machine interface, admittance controller was used to actuate the omni-directional platform and the guidance force was provided by the mobile base. In 2009, Allemand et. al. developed WalkTrainer which implemented a leg and pelvic exoskeleton on a holonomic mobile base. WalkTrainer featured a six degree-of-freedom pelvic orthosis that is fully actuated by six linear modules. Each linear module consists of a linear motor, position sensor, and force sensor. This module is controlled to change the compliance of the robotic system for limiting excessive interactive forces. Later in 2014, NaTUre-gaits [63] presented the device with a similar structure consisting of pelvis and leg exoskeletons connected to a mobile base. This walker has three degrees-of-freedom pelvic module on the right and left side of the frame and each module is controlled to guide the pelvis to follow a desired trajectory. Another walker to guide the pelvis was designed for correcting relative motion between pelvis and hip with a holonomic mobile platform. This robotic walker has redundant active degree of freedom that is set to control exoskeleton module and mobile base separately for ensuring transparency of the system. The mobile base moves to keep the pelvic center in the middle of the frame, and an impedance controller of the exoskeleton fully corrects abnormal pelvic motion of patients.

Recently, researchers have begun to extend their interest in designing functional gait training methods with pelvic module, e.g. for weight shift or balance, rather than designing kinematic structures of the pelvic module. Figure 1.2(h) [42] shows a simple lateral pelvic device using two cables on the side of pelvis, which is designed to assist weight transfer between the two legs. The cables provide a lateral pull in the direction of the stance foot while walking on the treadmill, where potentiometers on the ankle keep detecting phase of gait cycle. This device was tested with stroke patients who had difficulty to shift the weight in the lateral direction. And the pelvic displacement became more symmetric. Another pelvic device called Balance Assessment Robot (BAR) was developed by Olensek [81] in 2016 to provide perturbations during overground walking. The pelvic module is actuated to apply perturbations in three different directions (anterior-posterior displacement, media-lateral displacement, and transverse rotation) that are used to assess the balance responses of perturbations in transversal plane. This pelvic device is operated by an admittance controller in transparent mode so that the natural movement of pelvis is not affected unless a perturbation is applied.

As an extension of these studies, in the current dissertation, three different kinds of pelvic interventions are suggested with in-depth understanding of biomechanical response and gait adaptation to different correctional or therapeutic forces on the pelvis.

1.3 Tethered pelvic assist device (TPAD) as cable-driven robotic system

By virtue of unique advantages that TPAD offers, TPAD was chosen to explore different control methods to provide intervention at the pelvis in the present dissertation. (i) TPAD does not add inertia to the human body. TPAD only adds 320 g on the subject, which is 10 to 20 times smaller compared with other devices [111]. Added inertia on the child's body can result in undesirable motion during gait such as reduced anterior-posterior acceleration of the center of mass [70]. (ii) TPAD does not add rigid links on the human body. This feature allows free movement of the limbs with no restrictions on any degree of freedom (DOF) and reduces the therapist's effort to align the joint axes of the device with the human limbs or to adjust the size of the rigid links to the child's body. Because gait is a sequential coordination of limbs in three dimensions, restriction of specific DOFs can alter the entire gait pattern. (iii) TPAD imposes an overall external force during movement and allows the users to coordinate and control the limb DOF. Because abnormal gait patters are caused by impaired central nervous system [34], the type and level of lesion are quite different among individuals [22]. Instead of providing assistance to each joint with a predefined phase-dependent torque or displacement, TPAD promotes variability among joints and provides flexibility to the users in learning, as suggested by theories of human motor learning [65].

1.3.1 Hardware design

Tethered pelvic assist device (TPAD) is a cable-driven robotic system with actuated cables to apply three dimensional forces and torques as sketched in Fig. 1.3 [111, 112]. Figure 1.3 shows the different components of the TPAD. The motors and gearboxes (Kollmorgen, VA, USA) are set-up on an inertially rigid frame, from which the cables are routed through pulleys to connect to a fabric pelvic belt worn by a human subject. Each motor has a 4 N·m

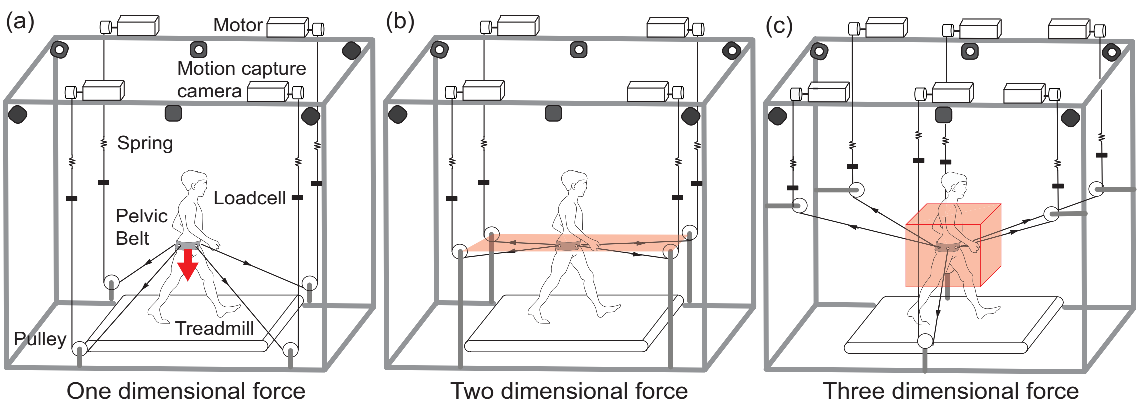


Fig. 1.3 Schematics of different configurations of tethered pelvic assist device (TPAD) for the purpose of applying (a) one, (b) two, and (c) three dimensional forces.

of continuous torque, and a cable reel of 5.08 cm (2 in) diameter is mounted on the gear box shaft to achieve a maximum continuous tension of 157 N in each cable. Load cells (tension sensors) are installed in series with each cable to measure the instantaneous cable tensions that can record up to a maximum of 890 N. Springs of stiffness 2.5 N/mm are installed in series with each cable to reduce the output impedance. Ground reaction forces (GRF) are recorded by force plates on a split belt treadmill (Bertec, OH). The controller is implemented with LabVIEW on a PXI real-time system (National Instruments, TX). While a subject walks on the TPAD, a 10-camera motion capture system (Bonita-10 series from Vicon, UK) is used as a part of the controller to track the human motion and cable attachment locations.

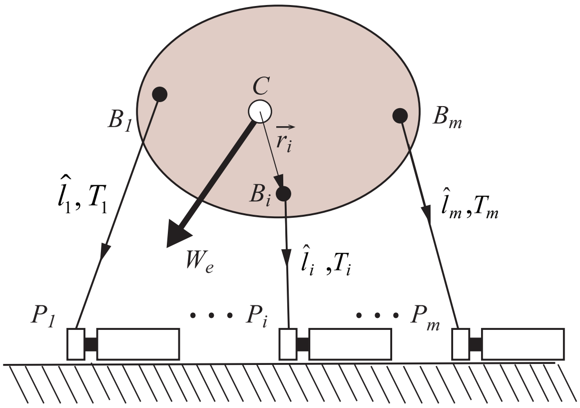
1.3.2 System model

TPAD is a cable-driven parallel system with actuated cables connected to the end effector, which is the human pelvis (Fig. 1.4). Each cable is modeled as a pure force at the attachment point. These cables together exert a wrench on the pelvis. Suppose $T \in R^m$ represents the tensions in the m cables, and $W_e \in R^n$ is the n-DOF external wrench on the pelvis, these are related to each other as

$$AT = W_e, \tag{1.1}$$

where $A \in R^{n\times m}$ is the structure matrix that depends on the system geometry and can be computed by the coordinates of the cable attachment points. For the case of 6-DOF wrench $W_e(n = 6)$ consisting of three-dimensional forces and moments, the matrix A is given by the following expression (Fig. 1.4):

$$A = \begin{bmatrix} \cdots & \hat{l}_i & \cdots \\ \cdots & \vec{r}_i \times \hat{l}_i & \cdots \end{bmatrix}_{6\times m} \tag{1.2}$$



P_i : attachment point of i^{th} cable on the frame
B_i : attachment point of i^{th} cable on the pelvis
$\hat{l}_i$: unit cable length vector
$\vec{r_i}$: vector from point C to i^{th} cable on the pelvis
T_i : tension of i^{th} cable
W_e : external wrench
C : point of action of wrench

Fig. 1.4 Cable-driven system illustration for constructing the structure matrix. The TPAD is a cable-driven parallel system with m actuated cables connected to the participant's pelvis having $n = 6$ DOFs. One end of each cable is attached to the pelvis, and the other end is attached to an electrical motor, shown as points B_i and P_i, respectively. These cables together exert an external wrench, W_e, on the pelvis.

where $\hat{l}_i$ is the i^{th} unit cable length vector from the end effector B_i toward the fixed routing point P_i, and $\vec{r}_i$ is the vector from the point of application C to the i^{th} cable attachment point B_i on the rigid body.

1.3.3 Controller

The controller consists of two different modules. The high-level controller computes the desired tension using a tension planner, and the low-level controller achieves the desired tension with feedback and feedforward terms. The low-level controller implements the desired cable tension at 1000 Hz using a force mode control scheme. An open-loop reference feedforward (T_{FF}) term was computed based on the desired tension value, and a feedback (T_{FB}) term was calculated from a proportional-integral-derivative controller. The motor constants (M_c) are empirically found for each motor and has the following values: motor 1

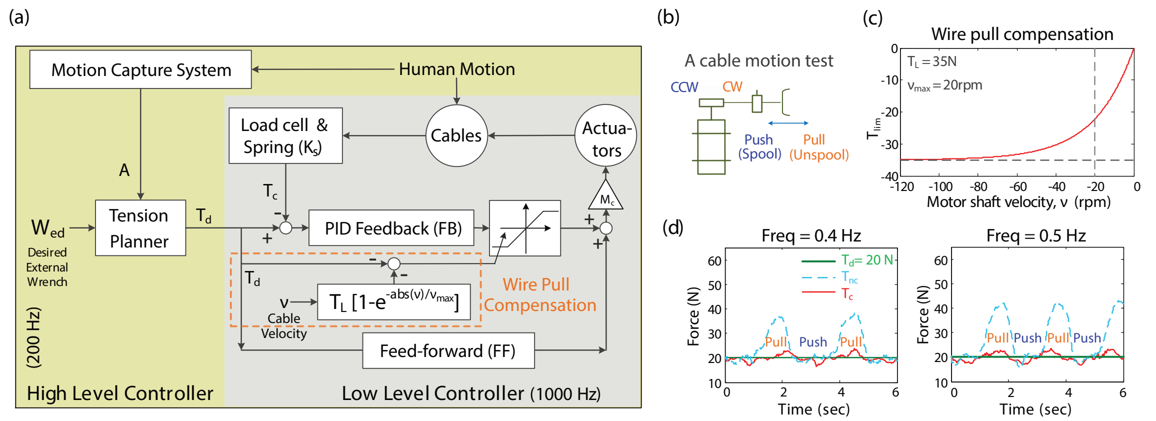


Fig. 1.5 TPAD controller schematic and testing for wire pull compensation (WPC). (a) Control architecture of low- and high-level controller. (b) Schematic of cable motion test. (c) WPC function. (d) A cable was pulled and pushed about 1 ft by hand at frequencies of 0.4 and 0.5 Hz for tension $T_d = 20$ N. T_c and T_{nc} are the cable tension values with and without the WPC term. The performance of controller is improved during pulling phase, when WPC is added.

(18.69 N/V), motor 2 (18.46 N/V), motor 3 (18.83 N/V), and motor 4 (18.89 N/V). Because the feedback term can have negative values, the minimum output of the feedback term was chosen to satisfy Eq. (1.3) to prevent cables from slacking:

$$T_{\mathrm{FB}} + T_{\mathrm{FF}} > 0. \tag{1.3}$$

However, the dynamics of the human and the actuators add errors in the cable tensions, as demonstrated in Fig. 1.5(d). To account for this phenomenon, wire pull compensation (WPC) was implemented which uses a relaxation of Eq. (1.3). It changes the lower limit $(T_{\mathrm{FB}})_{\mathrm{min}}$ for each motor as a function of the cable velocity, $v(t)$, as shown in Eq. (1.4).

$$\begin{aligned}(T_{\mathrm{FB}})_{\mathrm{min}} &= -T_{\mathrm{FF}} + T_{\mathrm{lim}} \\ &= -T_{\mathrm{FF}} - T_{\mathrm{L}}\left(1 - e^{\frac{-\|v(t)\|}{v_{max}}}\right),\end{aligned} \tag{1.4}$$

where T_{L} and v_{max} are two positive constant parameters. These parameters were tuned during a human walking test to achieve a responsive controller. T_{lim} is zero when the participant is stationary, that is, $v = 0$, and approaches $-T_{\mathrm{L}}$ exponentially as the negative cable velocity magnitude increases. T_{L} and v_{max} are constant values set by repeated human testing, respectively, $T_{\mathrm{L}} = 35$ N and $v_{\mathrm{max}} = 20$ rpm. The controller performance improved significantly during the cable-pulling phase, which is almost similar to the cable-pushing phase as shown in Fig. 1.5(d).

A tension planner computes desired tension of each cable to achieve the targeted wrench in the high-level controller. Cables in a cable-driven robot can only apply a pulling force on the end effector, and therefore, positive cable tensions must be maintained to retain control. For a general case of n-DOF system, at least $n+1$ cables are required for generating a desired wrench, $W_e \in R^n$ [72, 77]. Because the number of cables m is greater than the number of DOFs of the wrench n, Eq. (1.1) is underdetermined. Because this leads to infinitely many tension solutions, optimality criteria are needed to find a feasible solution set. Hence, a quadratic programming– based optimization scheme with equality and inequality constraints is implemented as below:

$$\min \left[\frac{1}{2}(T - T_p)^T (T - T_p)\right], \tag{1.5}$$

$$s.t. \qquad AT = W_e \quad \text{and} \quad T_{min} \leq T \leq T_{max}.$$

where $W_e = [F_x, F_y, F_z, M_x, M_y, M_z]^T$ consists of force and moment in three dimensions; x is medial-lateral direction, y is anterior-posterior direction, and z is the vertical direction. T_p is the optimized tension value of previous time step to obtain a continuous tension profile. T_{min} and T_{max} are the lower and upper bounds on the cable tension values, respectively.

1.4 Open and closed loop control: considering human performance

Section 1.3 describes the hardware and basic operating mechanisms of TPAD. In order to use the TPAD for rehabilitation purposes, it is further required to design the nature of force

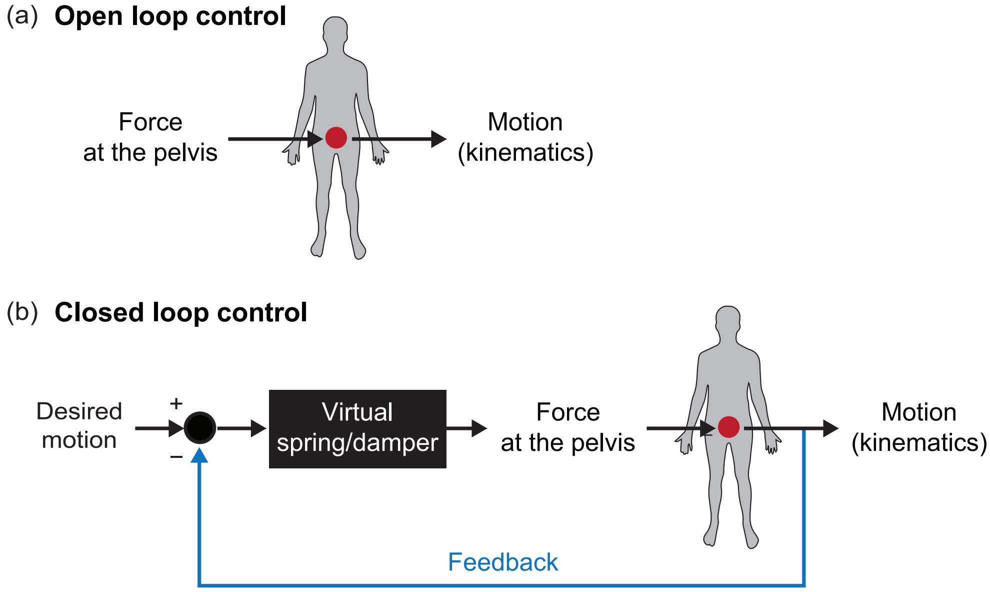


Fig. 1.6 Schematics of (a) open and (b) closed loop force controls on the pelvis. (a) Open loop method applies force regardless of the biomechanical information from the subject (b) while closed loop control interactively uses it to update the force based on the desired motion.

intervention. Control methods can be categorized by open and closed loop controls depending on whether the current state of the subject is involved in the control algorithm, as shown in Fig. 1.6. In open loop control, a device applies a pre-defined force profile, regardless of the biomechanical information of the subject (Fig. 1.6). The advantage of the open loop control is that it is simple to design as it does not need to measure the current status of the subject. As an example, in Chapter 3, a constant downward force is applied as a resistive training which strengthens legs of cerebral palsy children [48]. The magnitude and direction of this force was pre-defined and the controller does not adjust the characteristics of the force during training.

Unlike the open loop control method, the closed-loop control method applies a force in response to the performance or the current state of the subject, as in the Fig. 1.6(b). As the current state of the subject, the pelvic position is monitored continuously during the training and the force to be applied on the subject is determined based on the subject's pelvic motion [49]. The performance was evaluated by the error between the target pelvic center trajectory and the current pelvic center position. The virtual spring in the controller computes a force proportional to this error, normal to the target trajectory, in order to guide the pelvic motion closer to the target pelvic trajectory. This controller provides a force depending on the subject's performance and this concept is based on the fundamental paradigm of physical therapy, called 'assist-as-needed' [16]. This means that physical therapists typically pay attention to the motion of the patient's body, and then provide assistance only if the patient's motion deviates from the nominal gait pattern. This method encourages the voluntary movement of the patient and the physical therapist only provides verbal/haptic guidance when needed [33]. This assist-as-needed control method has been recently implemented in various exoskeletons for upper or lower limbs [16, 10, 33]. Chapter 2 demonstrates the closed-loop controller implemented in the cable actuated pelvic system (TPAD) and discusses its performance during treadmill walking.

1.5 Overview of the control approaches in the present dissertation

In order to correct the abnormal pelvic motion while improving the drawbacks of the previous designs, a cable-driven robotic system called tethered pelvic assist device (TPAD) [4, 110] has been recently developed as detailed in the Section 1.3. TPAD has the capability to apply three dimensional forces on the pelvis at the desired phase of the gait cycle, allowing a variety of interventions that apply different directional forces at different phases of the gait cycle. Also, this cable-actuated system has a distinct advantage of applying forces only at targeted point (pelvis), in contrast to rigid exoskeletons that restrict natural pelvic motion and add extra inertia due to the rigid linkages. The studies in the present dissertation suggest new control methods on how to design the therapeutic forces based on the understanding of human biomechanics during walking. Chapter 2 describes a new force type intervention based on the assist-as-needed controller to correct the abnormal pelvic movement during walking. In Chapter 3, downward pelvic pull is applied by cables in order to provide resistive forces and strengthen the antigravity muscles (soleus and gastrocnemius) of cerebral palsy children with crouch gait. Chapter 4 demonstrates the idea of modifying weight bearing by an asymmetric force field applied on the pelvis in the lateral direction. Finally, contributions of these works are summarized in Chapter 5 with suggestions for future work.

Chapter 2

On the adaptation of pelvic motion by applying 3-dimensional guidance forces using TPAD

As described in Section 1.4, there are two different control strategies depending on whether the biomechanical state of the subject is involved in the control algorithm (closed-loop control) or is absent (open-loop control). The present chapter illustrates the implementation of the closed loop method on tethered pelvic assist device (TPAD) in order to correct a pelvic motion while a human walks.

2.1 Assist-as-needed controller

In the early years, position controllers were used for different rehabilitation robotic systems to modify gait pattern by moving the pelvis of a patient through a predefined path so that it could be considered as a fully restricted method for controlling the position of pelvis [33]. This method was helpful for severely affected patients, but majority of patients needed more active involvement to improve their voluntary motor control. It is well known that the patients who kept completely passive during the training showed reduced EMG (electromyography) levels and metabolic cost during the training [101], which means that they are less actively involved in the training. In addition, these patients unintentionally experienced large resistance forces when they were trying to voluntarily move their lower limbs against the robot.

To overcome these drawbacks of position-controlled method, the idea of assist-as-needed controller has been recently applied to the robotic controller for rehabilitation purposes, which is motivated by physical therapy [33]. The principle of the assist-as-needed is to minimize the amount of assistance provided by physical therapists and encourage patients to participate during the training. This control method was implemented in different types of leg robotic devices to keep the leg motion within the predefined virtual tunnel [10, 33, 31, 16]. Banala et al. developed an active leg exoskeleton (ALEX) with assist-as-needed control which was successful to train stroke patients in order to regain the lower limb motor functions [10]. Also, Duschau-Wicke et al. conducted a pilot study for iSCI patients to walk in Lokomat with similar control strategy [31]. They reported that more active participation of patients were observed through sEMG for the assist-as-needed training than the position controlled trainings.

In this chapter, assist-as-needed strategy was adapted to correct and guide pelvic motion of subjects. The controller generates a target pelvic trajectory for the pelvic center and creates a guidance force that is normal to the target trajectory. The controller exerts a guidance force on the user's pelvis, depending on the pelvic center position with respect to the target trajectory, to assist the user in following the target pelvic motion. If the subject's pelvic center deviates normally further from the target trajectory, larger force is applied to the pelvis of the user. On the other hand, if the subject moves along the target trajectory properly, there isn't any force applied to the user. In other words, assist-as-needed controller provides force on the subject only when the controller detects the need to correct or guide the pelvic motion of the subject.

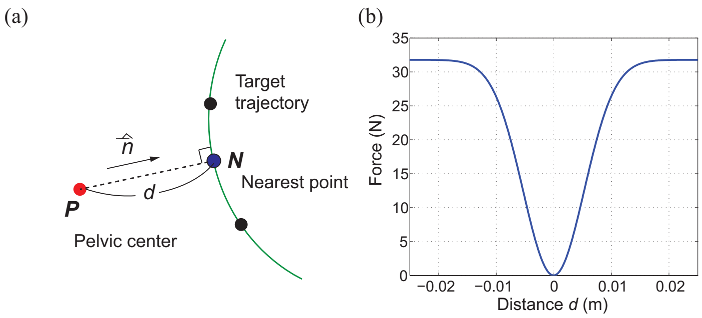


Fig. 2.1 (a) The pelvic target trajectory is presented by a solid green line, where a red dot denotes the current pelvic center point $\boldsymbol{P}$, and a blue dot denotes the nearest point $\boldsymbol{N}$. (b) The normal force profile $\boldsymbol{F_n}$ given by Eq. (2.1) using $K_n = 32\ N$ and $D = 0.0125\ m$.

2.1.1 Implementation of assist-as-needed control

Assist-as-needed controller is implemented by using force field controller. Force field controller is designed to generate a three dimensional guidance force along the target trajectory. This guidance force applies a correctional force on the pelvis, when the subject's pelvic center deviates from the target trajectory. As the subject's pelvis moves normally away from the target trajectory, larger guidance force is applied on the pelvis. The subject using this controller is supposed to experience a virtual tunnel that guides the pelvic motion along the target trajectory. The guidance force $\boldsymbol{F_n}$ is computed as follows.

In Fig. 2.1(a), let $\boldsymbol{P}$ be the current position of the pelvic center (red dot) which is computed as the mean of the left and right iliac crest markers. $\boldsymbol{N}$ is the closest point to $\boldsymbol{P}$ on the target trajectory (blue dot). The direction of the guidance force is along a vector from $\boldsymbol{P}$ to $\boldsymbol{N}$ which is normal to the target trajectory, denoted as a unit vector $\hat{\boldsymbol{n}}$. The magnitude of this force is defined as a function of the normal distance d between $\boldsymbol{P}$ and $\boldsymbol{N}$, as shown in Fig. 2.1(b). As the distance d increases, a larger normal force $\boldsymbol{F_n}$ is exerted on the pelvis. The equation for the normal force $\boldsymbol{F_n}$ is given by

$$\boldsymbol{F_n} = K_n \left[1 - e^{(d/D)^2}\right] \hat{\boldsymbol{n}}. \tag{2.1}$$

Table 2.1 Experiment modalities

Experiment		I	II	III-FF	III-VF	III-FFVF
Target Trajectory	Lateral ROM	Enl	Red	Red	Red	Red
	Vertical ROM	-	-	Red	Red	Red
	Vertical Height	-	-	Red	Red	Red
Feedback	Force	on	on	on	off	on
	Visual	on	on	off	on	on

FF: force feedback, VF: visual feedback, FFVF: force + visual feedback
Enl: enlargement, Red: reduction

A damping force $\boldsymbol{F_d}$ is added to the guidance force in order to minimize oscillations. The direction of this damping force is also along the vector $\hat{\boldsymbol{n}}$ and its magnitude is proportional to the velocity of the pelvic center $\boldsymbol{v}$, as

$$\boldsymbol{F_d} = -K_d\,(\boldsymbol{v}\cdot\hat{\boldsymbol{n}})\,\hat{\boldsymbol{n}}. \tag{2.2}$$

K_n, K_d, and D in Eqs. (2.1-2.2) are constant gains with $K_n = 32$ N, $K_d = 3$ N·s/m, and $D = 0.0125$ m. All constant parameters are determined by trial and error. K_n was chosen to be large enough to induce changes in the pelvic displacement and K_d was decided to limit the speed of oscillation [10]. The resultant force applied on the pelvis is the total force, $\boldsymbol{F}_{FF}$, which is

$$\boldsymbol{F}_{FF} = \boldsymbol{F_n} + \boldsymbol{F_d}. \tag{2.3}$$

2.2 Human Experiment

Thirty five healthy adults signed a written consent form approved by the Institutional Review Board (IRB) of Columbia University before the experiment. Each subject wore a pelvic belt with cables, reflective markers, and shoes with pressure pads. Markers were

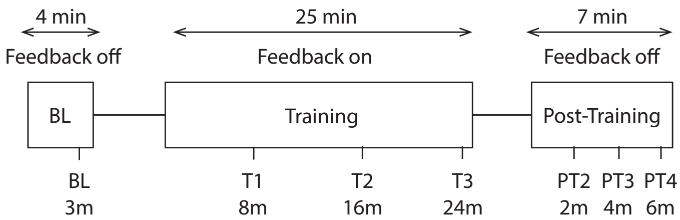


Fig. 2.2 The experimental protocol consists of baseline (BL), training (T), and post-training (PT) sessions. Force or visual Feedback is only provided during the training session.

attached on the right iliac crest, left iliac crest, sacrum, heels, and toes of the subject. During the experiment, all subjects walked at a constant treadmill speed of 3.8 km/h.

2.2.1 Protocol

The experimental protocol consisted of three sessions, as shown in Figure 4.1. During the baseline session (BL), subjects were asked to walk for four minutes and normal walking pattern was recorded. Data were collected in the last minute of this session. The Training session (T) was different for each subject depending on the experiment group the subject was assigned. Subjects walked with force or visual feedback for twenty five minutes. The target trajectory was designed by scaling up or down the baseline trajectory of each individual subject which will be detailed in section 2.2.3. Data were collected at every eight minutes and are denoted as T1, T2, and T3. For the post-training session (PT), all feedback was removed

Table 2.2 Subject information

Experiment	I	II	III-FF	III-VF	III-FFVF
Age (year) (year)	28.71 ± 3.12	25.86 ± 1.57	23.0 ± 4.51	28.57 ± 4.24	26.71 ± 4.89
Sex	7 Male	7 Male	6 Male 1 Female	6 Male 1 Female	6 Male 1 Female
Height (cm)	176.91 ± 7.71	173.57 ± 4.79	171.43 ± 5.53	181.00 ± 6.68	176.29 ± 7.18
Weight (kg)	71.43 ± 5.50	68.29 ± 11.47	61.57 ± 7.70	80.16 ± 11.04	71.29 ± 5.31

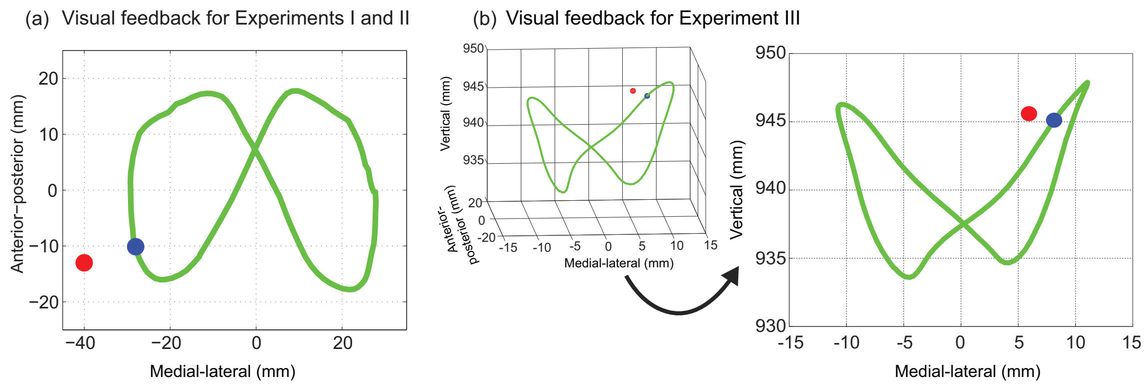


Fig. 2.3 Visual feedback is provided to the subjects with the pelvic target trajectory in a solid green line, where a red dot denotes the current pelvic center point, and a blue dot denotes the nearest point on the target trajectory. (a) In Experiments I and II, medial-lateral and anterior-posterior directional components of the trajectory are displayed. (b) Experiment III used the three dimensional trajectory to control the system, but the visual feedback provided a two dimensional display to the subject. Visual feedback showed only medial-lateral and vertical components of the trajectory.

and subjects were asked to walk for another seven minutes. Data were collected at the start, 2nd, 4th, and 6th minutes and are referred to as PT1, PT2, PT3, and PT4.

2.2.2 Experiment Modalities

Fourteen male subjects were divided into Experiments I and II as described in Table 2.2. During the training sessions in Experiments I and II, the force field was created in the transverse plane by using four cables attached to the belt. Simultaneously, visual feedback was also provided to these subjects while walking, as shown in Fig. 2.3. Subjects were instructed to follow the target pelvic trajectory and be aware of the forces applied at their pelvis. In Experiment III, twenty one healthy subjects were divided into three different groups (Table 2.1). Force Feedback (FF) group received the three dimensional guidance force from six cables attached to the pelvic belt. For the visual feedback (VF) group, a monitor in front of the subject showed target trajectory with the current pelvic center. The three dimensional target pelvic trajectory was projected in the coronal plane to show the lateral and vertical pelvic movement. Two dimensional trajectory was displayed, as it was more accessible and understandable to the subjects. The last group experienced both force feedback and visual feedback (FFVF) simultaneously until the end of the training.

2.2.3 Target trajectory design

The target pelvic trajectory was created from the recorded baseline data. This data was high pass filtered with a cut off frequency of 0.4 Hz to remove the slow translation over the treadmill platform. The filtered data was divided into gait cycles and the average value of ten gait cycles was used to generate the target trajectory for each individual.

For Experiment I, the target pelvic range of motion in the medial-lateral direction was enlarged by 40 % compared to BL as shown in Fig. 2.5 (a-b). In Experiment II, the target pelvic trajectory was reduced by 40 % compared to BL in medial-lateral direction as described in Fig. 2.5 (c-d). These changes in templates were chosen to provide challenging tasks to healthy subjects. The target trajectory was chosen such that it was beyond the range of baseline trajectory, but it is feasible for the subjects to walk without issues. For example, if the target trajectory of Experiment II was reduced excessively, subjects showed abnormal gait pattern to prevent legs from colliding with each other. For both target trajectories of Experiments I and II, pelvic range of motion in the anterior-posterior direction was kept the same as the baseline. The target pelvic trajectory was designed in the transverse plane, while allowing free movement in the vertical direction.

The baseline and target pelvic trajectory of Experiment III are displayed in Fig. 2.6 for a representative subject. This target trajectory was motivated from a comparison study of the CoM excursion of the healthy and cerebral palsy children [43]. In this study, healthy children presented smaller lateral and vertical CoM excursion than cerebral palsy children. In line with this study, (i) the target pelvic displacement in the lateral direction was reduced by 40 % from each subject's baseline, (ii) the vertical displacement was reduced by 60 %, and (iii) the mean height of the vertical pelvic movement was decreased in the target trajectory, so that the minimum values of the vertical component of both target and baseline trajectory were identical [43]. Similar to Experiments I and II, the anterior-posterior directional ROM was kept the same as the baseline.

2.2.4 Experiment set-up

In Experiments I and II, four cables were attached to the pelvis directed along a horizontal plane passing through the pelvis as shown in Fig. 2.4(a). The force field was created in this horizontal plane through the pelvis center. It is known in the cable-driven robot literature that the minimum number of cables that is needed to create a n DOF force-moment wrench is $n+1$ cables[77]. According to this rule, three cables are enough to generate a two dimensional force in the horizontal plane. But, in order to avoid having cables in close proximity of subjects and allow arm movements, four cables were used during the human experiment. The system set-up for Experiment III is shown in Fig. 2.4(b). In this experiment, since the pelvis was also guided in the vertical direction, cable configurations were changed to apply a three dimensional guidance force. Four cables were installed upwards and two cables were installed downwards. For the same reason as Experiments I and II, two extra cables were used even though a minimum of four cables was needed. These extra cables also helped to keep the moments applied to the pelvis small [112]. Extra wooden platforms were placed around the treadmill for safety.

2.2.5 Data analysis

All data were recorded for one minute and divided into gait cycles. A Gait cycle starts with the right heel strike on the treadmill and was calculated from the foot markers and the sacrum marker [124]. The divided data was time normalized to 100 % gait cycle. To validate the performance of the subjects and understand how they achieve the target trajectory, various parameters were estimated. First, ROM of the pelvic center was calculated for each gait cycle in all three directions. The ROM was computed by subtracting maximum and minimum displacements of the pelvis within a gait cycle. The data was filtered by a high pass filter

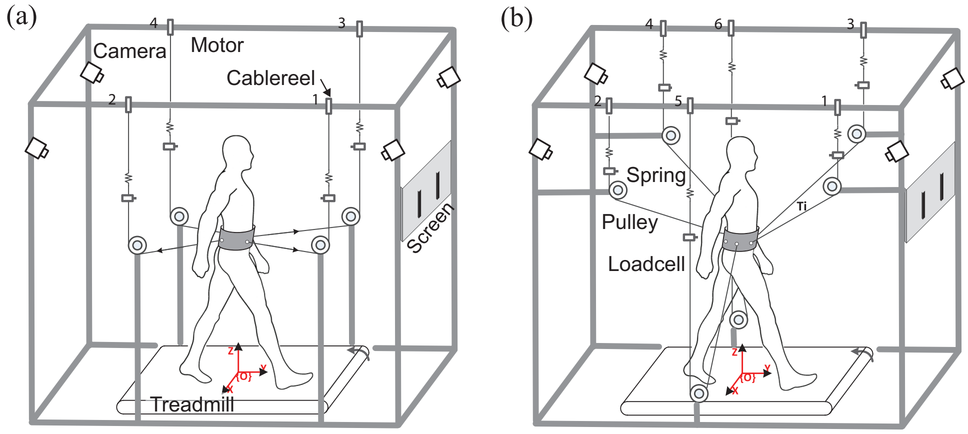


Fig. 2.4 (a) Schematic drawings of the system set-up for Experiments I and II. Four cables were attached to the pelvis directed along a horizontal plane passing through the pelvic center. (b) Set-up for Experiment III consisted of two downward and four upward cables. Global frame is set at the middle of the treadmill.

(cutoff frequency: 0.4 Hz) in order to eliminate the slow translation on the treadmill. The percentage error of ROM is defined by the difference between the each session's ROM and the target ROM, normalized by the baseline ROM as

$$\%\ error\ ROM = \left| \frac{target\ ROM - actual\ ROM}{baseline\ ROM} \right| \times 100. \tag{2.4}$$

Also, gait parameters were computed for each session. Step width was defined as the maximum medial-lateral distance between the right and left heel markers during the double support period after the heel strike. Step length for a leg was defined as the anterior-posterior distance between the heel markers of two legs at the moment of the leg's heel strike. Both step width and step length were averaged for right and left heel strike. These values were normalized by each subject‘s height. For each parameter, one minute data was averaged to compute the representative value. Statistical analysis was run separately two times to examine significant effects on training sessions and post-training sessions respect to the baseline session. Non-parametric repeated measure (Friedmans ANOVA) was performed

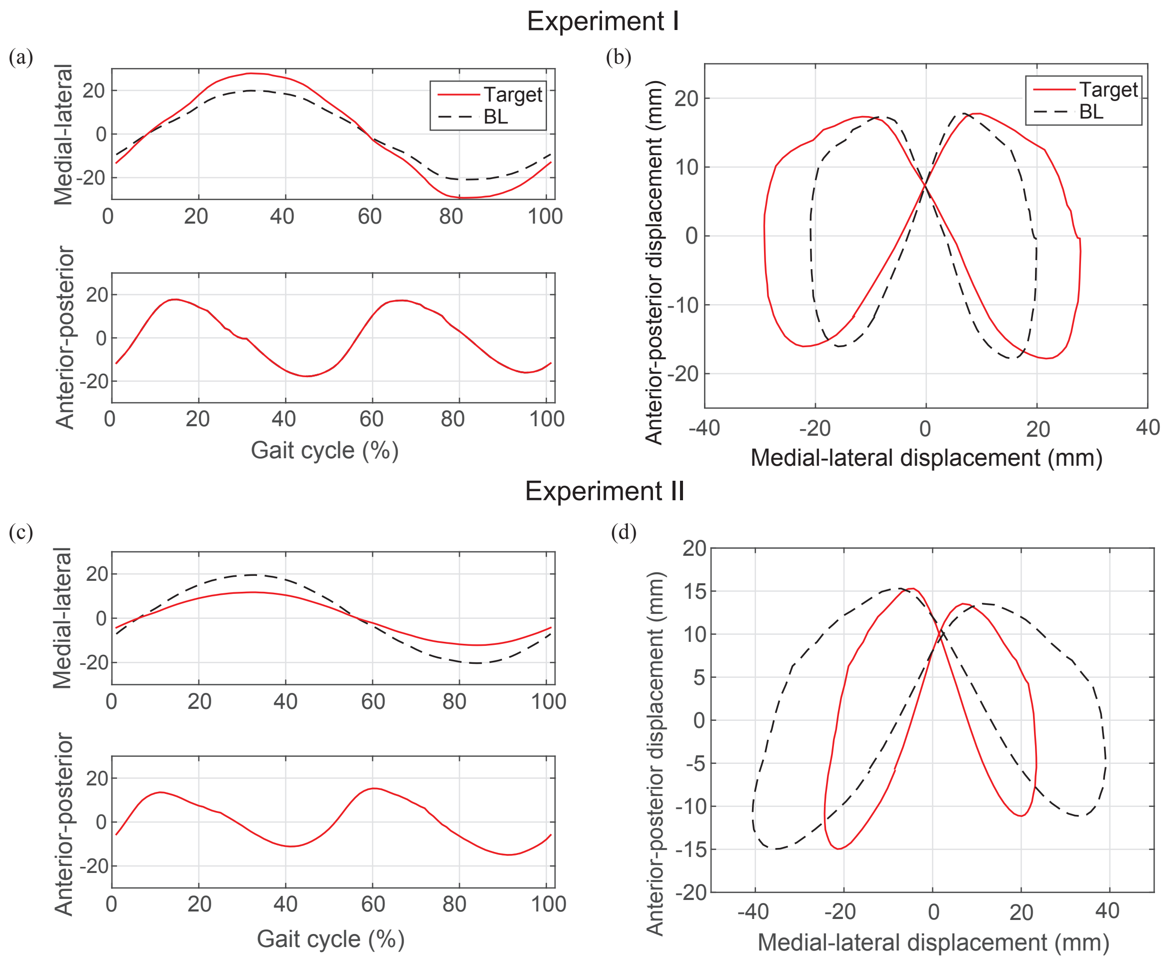


Fig. 2.5 Sample baseline (BL) and constructed target pelvic trajectories for Experiments I and II. The target pelvic trajectory (a-b) of Experiment I with 40 % increase in lateral range of motion (ROM) and (c-d) of Experiment II with 40 % reduction in lateral ROM.

with significance $\alpha = 0.05$ and Wilcoxon signed rank test was conducted for the pairwise comparison with Bonferroni correction [123].

2.3 Results

2.3.1 Experiments I and II

Representative pelvic trajectories for a subject are presented in the transverse plane for the baseline (BL) and post-training (PT) sessions in Fig. 2.7(a-b). Solid red lines in both figures

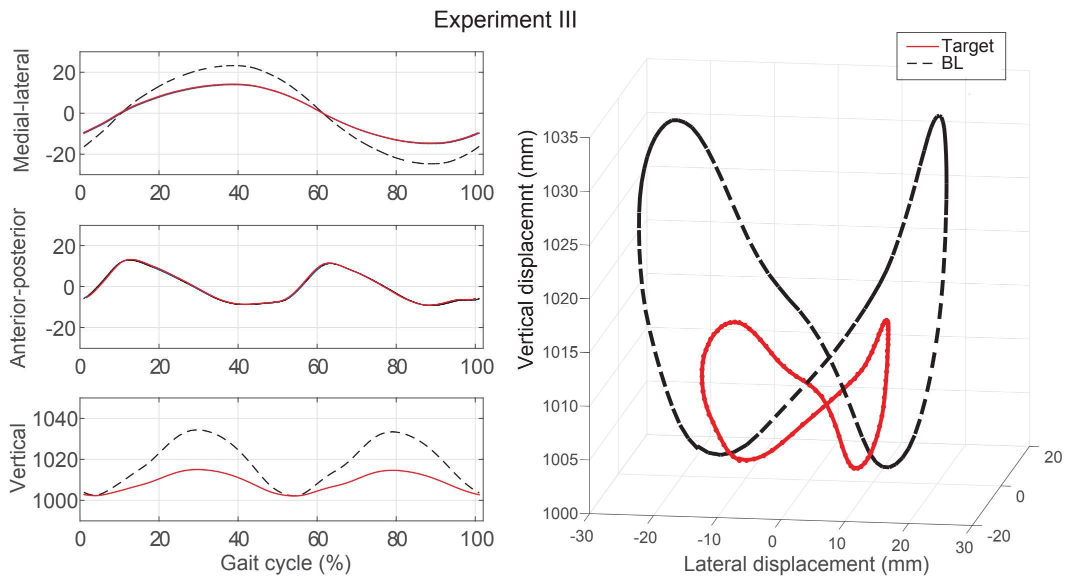


Fig. 2.6 Sample baseline (BL) and constructed target pelvic trajectory for Experiment III. The target trajectory shows the reduced lateral and vertical ROM when compared to the baseline trajectory.

display the target pelvic trajectory that is calculated from the baseline of each subject. The target pelvic trajectory has enlarged (Experiment I) or reduced (Experiment II) medial-lateral span compared to the baseline pelvic trajectory. In addition, blue lines show that the subject kept the increased (Experiment I) or decreased (Experiment II) lateral ROM during the post-training trials.

Percentage error of the lateral ROM for Experiments I and II are presented in Fig. 2.7(c) with mean and standard error values. The non-parametric repeated measure test reported significant changes for both experiments during post-training sessions ($p<0.05$). The post-hoc pairwise test showed significant changes between the baseline and the post-training values. The result with the increased target (Experiment I) showed significant decrease of the percentage error in BL-PT2 and BL-PT3, while the reduced target experiment (Experiment II) showed significant decrease of percentage error in BL-PT3. It indicates that subjects were able to retain the enlarged or reduced pelvic motion about seven minutes after the training. Gait parameters are also displayed in Fig. 2.8(a-b). The step width of Experiment I showed increase during training and post-training sessions. Wilcoxon signed rank test

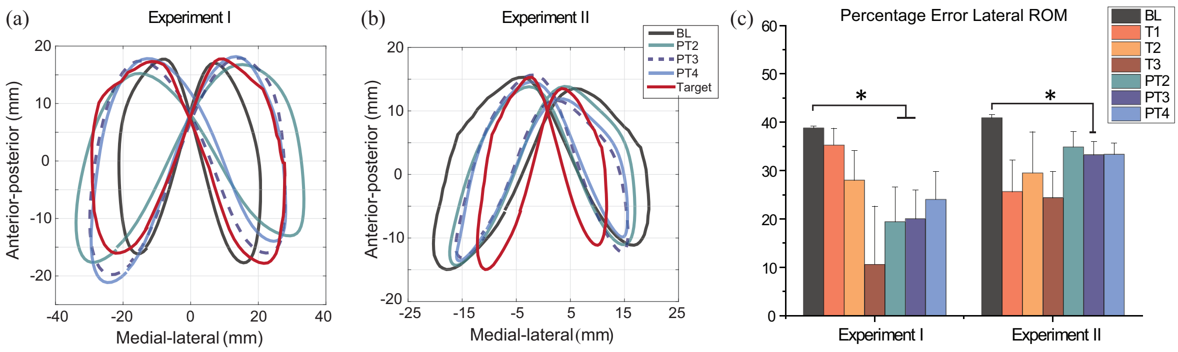


Fig. 2.7 The pelvic trajectory of the baseline and post-trainings in the transverse plane for (a) Experiment I and (b) Experiment II of a representative subject. Experiment I enlarges the target pelvic lateral ROM by 40 % from the baseline, while Experiment II decreases it with the same percentage. (c) The percentage error of the lateral ROM is shown for the baseline, training sessions, and post-training sessions. Both experiments show significant changes in the post-training sessions compared to the baseline ($\alpha = 0.05$). Significant pairs respect to the baseline are presented with different levels: * p<0.05, ** p<0.01, *** p<0.005.

reported significant increase during the training sessions for pairs BL-T1. In Experiment II, the step width decreased during the training and after the training. Wilcoxon's test reported significance in the post-training session of Experiment II (BL-PT2, BL-PT3, and BL-PT4). The other gait parameter shown in Fig. 2.8 is the step length. It showed increase for Experiment I for the post-training compared to the baseline. As significant pairs, BL-PT3 and BL-PT4 were reported. The step length of Experiment II showed decrease in the training session (BL-T1), but the post-training values returned to the baseline values. The last parameter of Fig. 2.8 is the anterior-posterior ROM of the pelvis, which significantly increased for Experiment I during the training and post-training. The pairwise comparison reported significant pairs BL-T3, BL-PT2, and BL-PT3. The anterior-posterior ROM of Experiment II showed similar trend in its step length values. Significant pair BL-PT4 were found in the anterior-posterior ROM data for Experiment II.

2.3.2 Experiment III

The results of Experiment III with the force feedback, visual feedback and the force plus visual feedback groups are presented in Figs. 2.9 and 2.11. For all three groups, the target

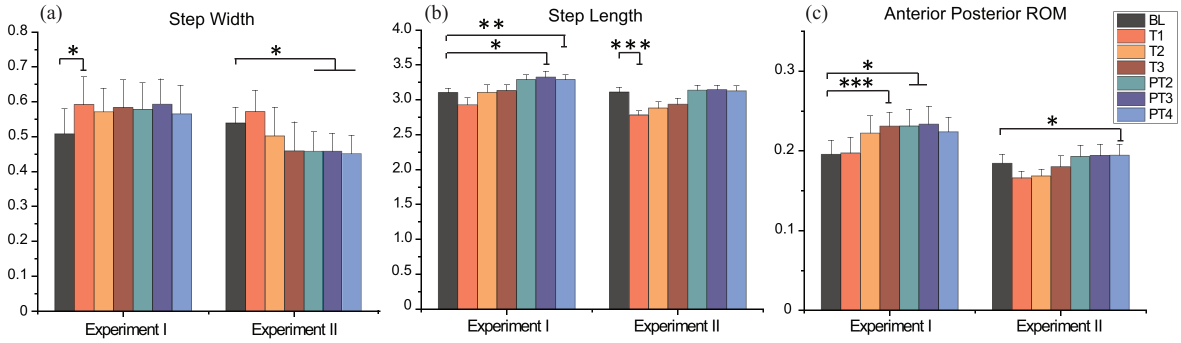


Fig. 2.8 (a) Step width, (b) step length, and (c) anterior posterior ROM of Experiments I and II. The subjects changed these three parameters to achieve the target pelvic trajectory. All data are scaled with the height of the subject.

pelvic trajectory was designed to reduce simultaneously the lateral ROM, the vertical ROM, and the vertical mean height of the pelvic trajectory.

Experiment III-FF

The Force Feedback (FF) group showed significant decrease in the percentage error of the lateral ROM during the training session (BL-T2 and BL-T3). Also, during the post-training, subjects were observed to retain the trained lateral ROM, but the Friedman's test reported no significance. The FF group also significantly decreased the vertical ROM during the training ($p<0.01$). After the force feedback was removed, subjects showed negative after-effects for the vertical ROM ($p<0.01$). For these after-effects, Wilcoxon pairwise test reported two significant pairs for the vertical ROM (BL-PT3 and BL-PT4). Lastly, the percentage error of the mean height showed slight decrease for the force feedback group, but no significant change was found for both the training and the post-training sessions. From above observations, it was found that force feedback had stronger effect on the subjects' pelvic trajectory to change the lateral and vertical ROM.

Gait parameters accompanying the above changes are presented in Fig. 2.11(a-b). For the FF group, step width was found to decrease significantly during the training ($p<0.01$). Wilcoxon's test reported significant pairs as BL-T2 and BL-T3. After the training, step width remained smaller than the baseline, but no significance was reported. The force feedback

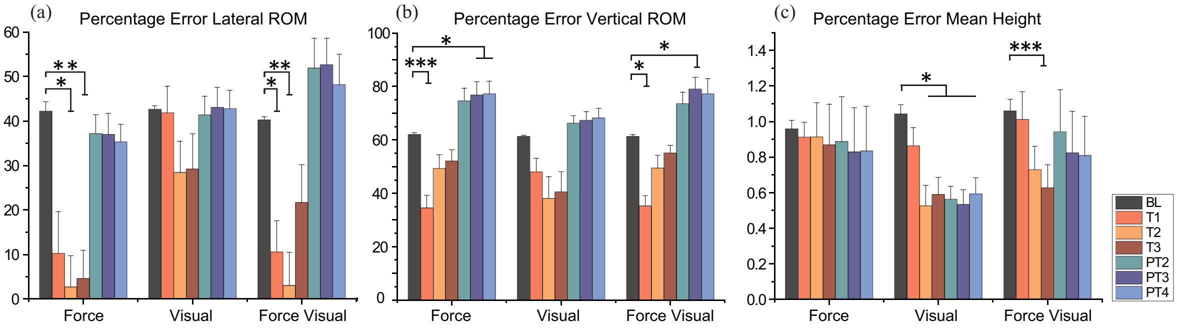


Fig. 2.9 The percentage error of (a) the lateral ROM, (b) the vertical ROM, (c) and the vertical mean height values are displayed for three different groups in Experiment III. The force feedback shows significant training effects on both the lateral and the vertical ROM, while the visual feedback presents significant change in the mean height. Significant pairs respect to the baseline are presented with different levels: * $p<0.05$, ** $p<0.01$, *** $p<0.005$.

group also showed decreasing trend in the step length. However, it was observed that there was no significant effect on the step length. The anterior-posterior pelvic ROM showed an increasing trend for both training and post-training sessions. However, for anterior-posterior ROM, the only significant pair was BL-PT3.

Experiment III-VF

In the visual feedback (VF) group, the medial-lateral ROM showed decreasing trend during the training, but no significant relation was reported for both training and post-training. The percentage error of the vertical ROM showed decreasing trend during the training and negative after effects. However, Friedman's test did not report any significance for all sessions. The percentage error of the mean height was reduced for both training and post-training sessions. During the training, significant pairs were reported for BL-T2 and BL-T3. Also, all three post-training sessions showed significant pairs compared with the baseline.

The visual feedback group showed similar trend as the force feedback group for the step width. The step width decreased during the training (BL-T2 and BL-T3) and the effect remained after the training (BL-PT3 and BL-PT4). Also, the step length decreased during the training and showed negative after-effect for post-training sessions. The anterior-posterior ROM decreased during the training, but it increased after the training compared to the

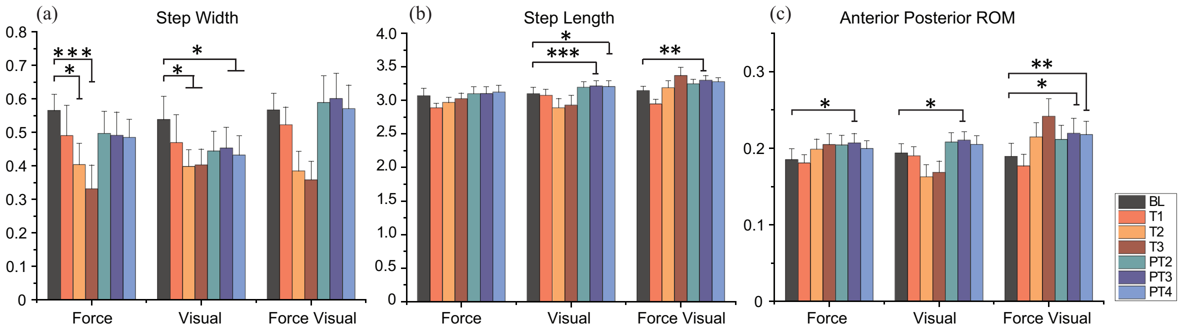


Fig. 2.10 (a) Step width, (b) step length, and (c) anterior-posterior ROM of Experiment III. All subjects show the tendency to alter these three parameters to achieve the target pelvic trajectory. All data are scaled by the height of each subject.

baseline. The Friedman's test reported significance only for the post-training sessions with the significant pair BL-PT3.

Experiment III-FFVF

The third group who received both force feedback and visual feedback (FFVF) showed combined effect of FF and VF groups. The percentage error of the medial-lateral ROM showed striking decrease during the training sessions. The non-parametric pairwise comparison reported significant pairs for BL-T1 and BL-T2. Also, the percentage error of the vertical ROM showed similar trend as the force feedback group. The training decreased the vertical ROM percentage error ($p<0.005$) and negative after effects were observed for the post-training session ($p<0.005$). The force plus visual feedback group also showed significant decrease in the mean height of the pelvic movement (BL-T3). After training, there was a trend to retain the decreased mean height, but no significance was reported.

The FFVF group showed decrease in the step width during the training sessions ($p<0.05$). But no pairwise comparison was reported. For this group, step length continuously increased during the training ($p<0.01$). After the training, step length retained the increased value with significant pair BL-PT3. The anterior-posterior ROM showed similar trend with the step length. The anterior-posterior ROM increased during the training and retained the increased value after the training (BL-PT3 and BL-PT4).

2.4 Discussion

2.4.1 Experiments I and II

The assist-as-needed control strategy was demonstrated to be capable of guiding and changing the subjects' pelvic movement. Subjects who participated in the experiment were able to move their original pelvic trajectory towards the target trajectory by reducing or enlarging the lateral ROM. The target trajectory with increased lateral ROM (Experiment I) increased subject's lateral ROM of the pelvic movement, while the target with reduced lateral ROM (Experiment II) was able to reduce subjects' lateral ROM of the pelvic movement. The alteration in pelvic movement is retained during the post-training sessions for about seven minutes. The results of the current work demonstrate the capability of the proposed system as a pelvic device to correct pelvic movement.

While forces were applied during the training session, trends in the changes were found to be different between Experiments I and II. In Experiment I with enlarged lateral ROM, subjects tended to gradually change the pelvic trajectory towards the target. While in Experiment II with reduced lateral ROM, subjects showed immediate changes during the training (Fig. 2.7 (c)). This would indicate that for Experiment I, the forces acted mainly in feedback so that the subject were able to learn the new target trajectory by understanding the nature of the force field applied on the pelvis. As the training went on, subjects may have developed a better understanding of this force and put conscious effort to change the pelvic movement until they reached the target trajectory. However, in Experiment II, the force seemed to play a different role to reduce the lateral ROM instead of feedback. Forces that are proportional to the error can be considered as a virtual spring, which help to stabilize the lateral pelvic movement. From the perspective that human gait is passively unstable in the lateral direction [57], stabilizing components such as a virtual spring/damper will reduce the ROM immediately, as observed in Experiment II. This observation is similar to the work of Donelan [30], in which a physical spring was installed in the lateral direction to stabilize the pelvis.

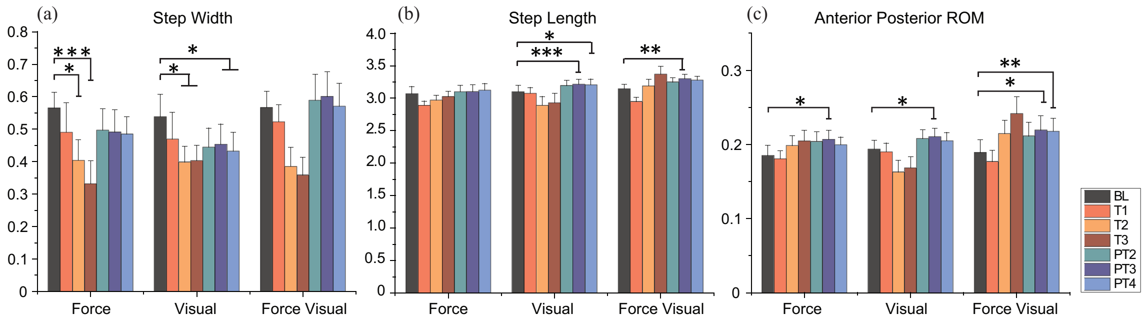


Fig. 2.11 (a) Step width, (b) step length, and (c) anterior-posterior ROM of Experiment III. All subjects show the tendency to alter these three parameters to achieve the target pelvic trajectory. All data are scaled by the height of each subject.

Even though there was no direct intervention on the lower extremities during the training, subjects changed their gait parameters to be compatible to the target pelvic trajectory. Subjects in Experiment I who walked with enlarged lateral target showed larger step width, whereas subjects in Experiment II with reduced target pelvic trajectory presented smaller step width than their baseline session (Fig. 2.8 (a)). Subjects in Experiment I may increase their step width in order to increase their base of support to keep balance while walking [64]. They increased their base of support in the lateral direction to keep the center of mass within the base of support due to the increased lateral pelvic movement. Other subjects in Experiment II decreased their step width as the virtual spring from the controller helped the subjects to stabilize the lateral pelvic movement. Step width is decided as a trade-off in between the base of support that is needed to stabilize the body and the energy expenditure required to change the direction of CoM in lateral direction [29]. As the virtual springs helped to stabilize the body, subjects tended to decrease the step width and take advantage of lower energy cost. This observation is consistent with the biomechanical model of Zijelstra et. al. [125] that shows a proportional relation between the step width and the lateral pelvic displacement. In this biomechanical model, step frequency is also included as a minor factor that affects the lateral pelvic movement. This explains why T1 in Experiment II decreased the percentage error lateral ROM (Fig. 2.7(c)), even though the step width was increased (Fig. 2.8(a)). Decreased lateral pelvic displacement in T1 was caused by significantly increased stride frequency, or namely the decreased step length (Fig. 2.8(b)) while walking with constant speed [17, 125]. Subjects who changed the step length presented proportional trend in the anterior-posterior ROM, since anterior-posterior pelvic displacement also depends on the stride length [125].

2.4.2 Experiment III

Subjects who participated in Experiment III were able to learn the target trajectory during the training. However, it was observed that subjects did not retain the training effect of the medial-lateral ROM after the training. This may be caused by the multiple tasks that impede the cognitive processing during training and leads to possible slowing of the learning process [45]. Experiment III changed three different pelvic parameters and subjects had more challenges to learn or retain the new target trajectory. Simple tasks such as Experiments I and II seem to be more efficient to learn and retain a new pelvic trajectory.

For the vertical ROM in Experiment III, the vertical ROM showed significant after-effect instead of retention. This would be caused by the fact that the percentage error was larger in vertical ROM during training (Fig. 2.9(b)), in which there was always a downward force applied on the subject. Even though the downward force was not there during the post-training sessions, subjects kept the strong push-off to resist against the downward force during the training. Subjects changed the feed-forward motor commands while they were walking in the device [58]. When the cables were removed, the absence of downward force excessively accelerated the pelvis upward which may lead to the larger vertical ROM.

Three different groups were recruited for Experiment III to understand the role of force and visual feedback in Experiments I and II. Even though subjects in Experiments I and II anecdotically commented that the visual feedback was not helpful, the contribution of the different feedback in learning needed investigation. During the training, it was observed that the force feedback group showed immediate changes in the pelvic trajectory than the visual feedback group. This is because the applied force using the cable-driven robot provided a haptic guidance to the pelvis and directed the subjects towards the target trajectory. On the other hand, the changes in the visual feedback group were gradual. The visual feedback group put conscious effort to follow the target trajectory which involved more explicit learning. We believe that a paradigm using a combination of both visual and force feedback to add both the conscious and the haptic guidance could strengthen the training effect. Kim et. al. [54] also reported that a combination of visual guidance and force constraints is successful to adapt a new gait pattern with a robotic exoskeleton.

The other noticeable difference in these groups was that the force group focused more on the lateral pelvic movement, while the visual feedback group focused more on the vertical movement. This might be caused by the fact that forces applied in the lateral direction have larger impact on the CoM than in the vertical direction. Considering human body as an inverted pendulum, the lateral force can apply larger torque on the support foot than the vertical forces. In addition, subjects in the visual feedback group paid more attention to the vertical pelvic movement. For complex task such as Experiment III, visual feedback and force feedback showed mutual supplemental roles to learn a new target trajectory. The combined effect of both feedbacks was well presented in the force plus visual feedback group which showed changes both in the lateral and vertical pelvic movement.

2.5 Conclusion

The present work demonstrates the capability of a novel control method to guide and correct pelvic movement. With this new control method, subjects were able to learn a new pelvic trajectory and retain the training effect. Even though there was no direct intervention on the lower extremities, the movements of the legs also changed in order to be compatible to the target pelvic trajectory. In addition, we found from the results of Experiment III that a training paradigm using both visual and force feedback could enhance the training effect. Instead of providing intervention at the lower extremity such as using exoskeleton type devices, TPAD and its controllers can directly intervene at the center of mass to improve impaired gait of patient groups and provide better mobility which is crucial to their quality of life.

Chapter 3

Robot-driven downward pelvic pull to improve crouch gait in children with cerebral palsy

Chapter 2 described the novel assist-as-needed controller to guide a pelvic motion in three dimensions, which is based on a closed-loop method. This chapter introduces a clinical study of employing TPAD based on the open-loop method. This study was performed to improve the crouch gait that is found among children with cerebral palsy.

3.1 Crouch gait and rehabilitation methods

In the United States, 3.6 children out of every 1000 school-aged children have a diagnosis of cerebral palsy (CP) [120]. These children show abnormal gait patterns, for example, jump gait, equinus gait, and crouch gait [105, 92]. The crouch gait is characterized by excessive flexion of the hips, knees, or ankles [105, 34, 14]. Children with crouch have slow walking speed, reduced range of motion of the joints (ROM), small step length, large body sway, and absence of a heel strike [80]. In addition to other complications, crouch gait increases the energy cost of walking, causes pain, and results in joint degeneration over time [102].

Crouch gait is caused by weak extensor muscles that do not produce adequate muscle forces to keep upright posture under gravity [104, 102]. Among the extensor muscles, the soleus plays an important role to prevent knee collapse during the middle of stance phase [85]. The soleus muscle keeps the shank upright during the mid-stance phase of the gait to facilitate extension of the knee. When dorsiflexion of the foot increases by weak soleus muscles, the upper part of the shank leans forward, and this contributes to the flexion of the knee. In this posture, the body weight creates a larger flexion torque on the knees, which leads to collapse of this joint [34]. The soleus is also responsible for propulsive forces on the human body during midstance to terminal stance phase. The activation timing of the muscles with respect to the gait cycle is also an important issue. If this timing is not appropriately synchronized with the gait, it results in poor kinematics of the leg and ground reaction forces (GRFs). If two muscles with opposite functions activate together, muscle efficiency is reduced, which leads to early fatigue. Coactivation of plantar flexor muscles and quadriceps is a major issue in crouch gait. During normal gait, plantar flexor muscles accelerate the center of mass forward during the late stance phase, whereas the quadriceps decelerate the center of mass during the early stance phase [104]. Children with crouch gait are often observed to have earlier activation of gastrocnemius and soleus muscles, and this contradicts the function of quadriceps muscles, contributing to inefficiency of crouch gait [104]. Dietz and Berger have also reported that plantar flexor muscles are impaired in CP children, showing lower amplitudes and poor timing, compared with healthy peers [26].

Surgery is often recommended for children with CP to lengthen the tightened flexor muscles. However, previous results show that this does not improve walking because extensor muscles still remain weak [102]. Although early literature suggested that muscle strengthening of the legs can worsen spasticity and coactivation in the muscles, more recent research disputes these claims[24, 7]. Different strength training strategies have shown increased muscle strength without worsening spasticity or muscle coactivation [7, 15, 97, 73]. Currently, different resistive strength training schemes are used in clinical practice with leg presses, rubber bands, or loaded sit-to-stand to strengthen the extensor muscles. Whether the effects of strength training translate to overground walking still remains under debate [98, 96]. One possible reason suggested by researchers is that the task during the training is not directly related to walking. Another method often used in clinical practice to retrain walking in children with CP is treadmill training with partial body weight suspension [95, 12, 27]. However, no studies with partial body weight suspension report postural correction of the crouch gait.

3.2 New rehabilitation method with a cable actuated pelvic device

In this study, we took an approach opposite to what is used in conventional therapy with these children. Instead of partial body weight suspension during treadmill walking, participants are trained to walk with a force augmentation. The scientific rationale behind this study is to strengthen the soleus muscles, which is a major pathological cause for the crouch gait. It was previously shown that soleus muscles are activated more strongly among the lower limb muscles when extra weight is added on the human body during gait [68]. This is because the soleus is the major weight-bearing muscle during the single stance support. From these observations, our study is designed to apply an additional downward force on the human body to intensively retrain the activity of the soleus muscles. A downward force equivalent to 10% of the body weight was chosen on the basis of the results of healthy children carrying backpacks—this was the minimum weight to show notable changes in posture or gait parameters during walking[20, 61].

A cable-driven robot called Tethered Pelvic Assist Device (TPAD) [111] was programmed to apply a downward pull force through the center of the pelvis while the participants walked on a treadmill. TPAD consists of a lightweight belt worn by a participant on the pelvis to which several wires are attached. The tension in each wire was controlled in real time by a motor placed on a stationary frame around the treadmill, based on real-time motion capture data from cameras. TPAD is a unique cable-driven robotic device because it applies external forces on the human body during motion, and the training with this device is distinctive because it provides necessary strength and coordination training while walking.

Six children diagnosed with CP and exhibiting crouch gait underwent 15 training sessions of 16 min each, over a duration of 6 weeks. We hypothesized that, as the study participants raise their center of mass against the applied downward force from TPAD during walking, they will improve control of their ankle plantar flexor muscles, especially the soleus, and improve their crouched posture. The muscle strength and coordination were investigated using electromyography (EMG) data from the first and after last sessions of training. Kinematics and GRF were monitored continuously throughout the training.

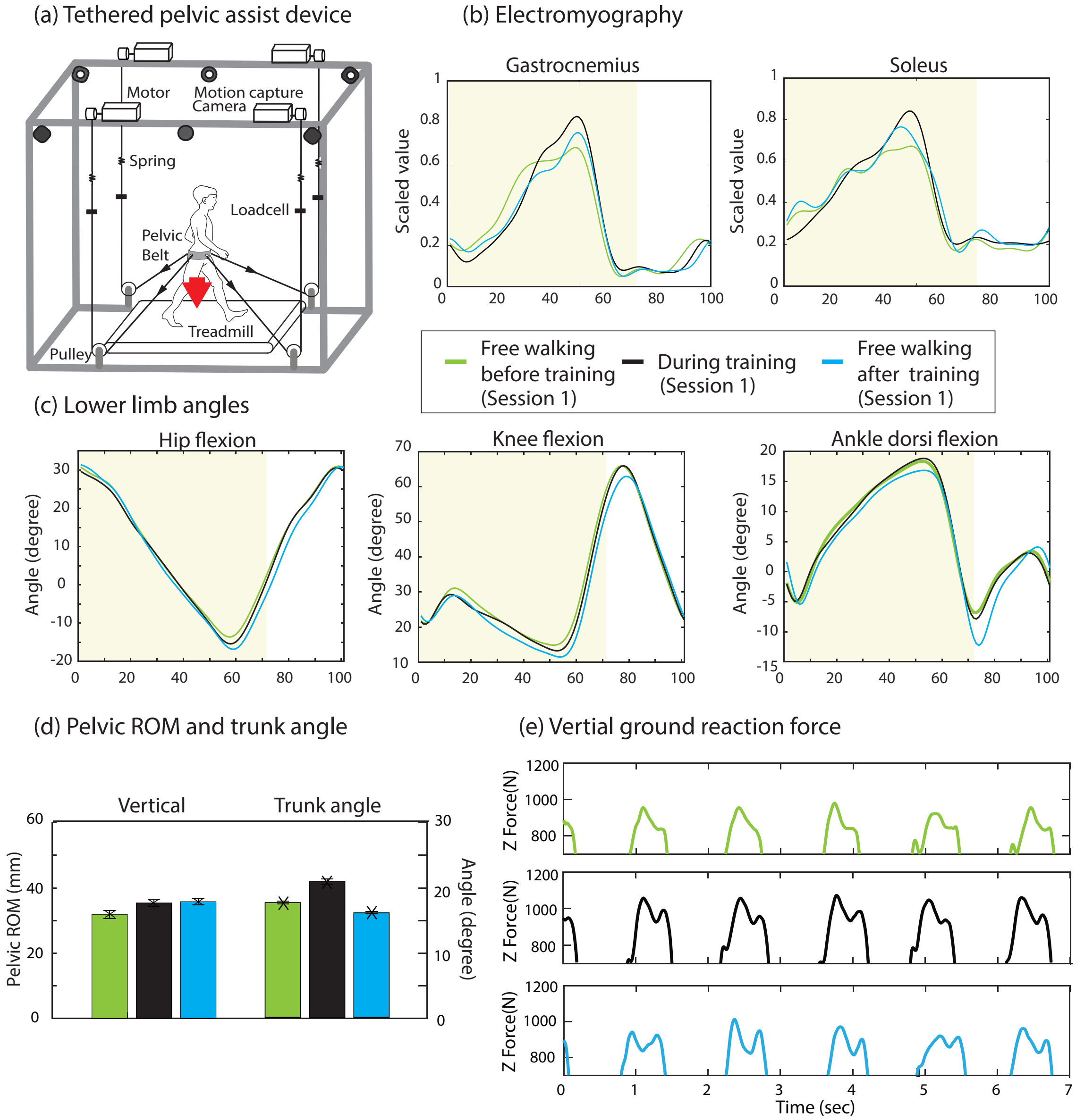


Fig. 3.1 System setup and results of a representative participant in a single training session. (a) TPAD system setup. (b) EMG data, (c) lower limb angles for hip, knee, and ankle, (d) pelvic translational ROM and trunk angle, and (e) vertical GRFs during session 1 including free walking before training, during training with force, and free walking after training. Data during training (black line) were recorded at the seventh minute of training, and data after training (blue line) were recorded at the third minute after training. Yellow shaded area presents the stance phase of the gait cycle. Trunk angle is defined by two markers attached to the seventh cervical vertebra and sacrum, averaged over the gait cycle.

3.3 Human experiment

3.3.1 Experiment protocol

Each participant participated in 3 training sessions per week for a total of 15 training sessions. For each training session, participants walked on a force plate–instrumented treadmill for roughly 30 min, with body reflective markers recorded by a motion capture system. During the first training session of each week, the treadmill speed was decided before the training. The treadmill speed was selected by increasing the speed slowly in increments of 0.1 m/s until the child had difficulty walking. Then, the speed was slowed down until the child felt comfortable to walk at that speed. The study protocol consists of free walking before training, training with TPAD, and free walking after training (Fig. 3.2). Free walking was recorded without tethers attached to the pelvic belt. After 5 weeks of training, participants visited again during the sixth week to characterize their walking on the treadmill with the same speed as the first week. One week before the first training session and 2 weeks after the last training session, the Six Minute Walk test was conducted to measure overground walking speed. Besides walking test, BBS and TUG tests were also performed. EMG was recorded only for the 1st and 16th sessions while walking on the treadmill at the same speed. The study was approved by Columbia University's institutional review board.

3.3.2 Participants

Six CP children with crouch gait participated in the robotic training with TPAD. These children were diagnosed as non-toe walkers, because the present study is focused on the crouch gait. Participants ages 20 years and below were recruited and classified as GMFCS level II. Exclusions criteria were (i) Botox injections within the past 3 months, (ii) ethanol injections within the past 6 months, and (iii) dorsal rhizotomy surgery within 1 year before the study. Detailed information of each participant is shown in table 3.1.

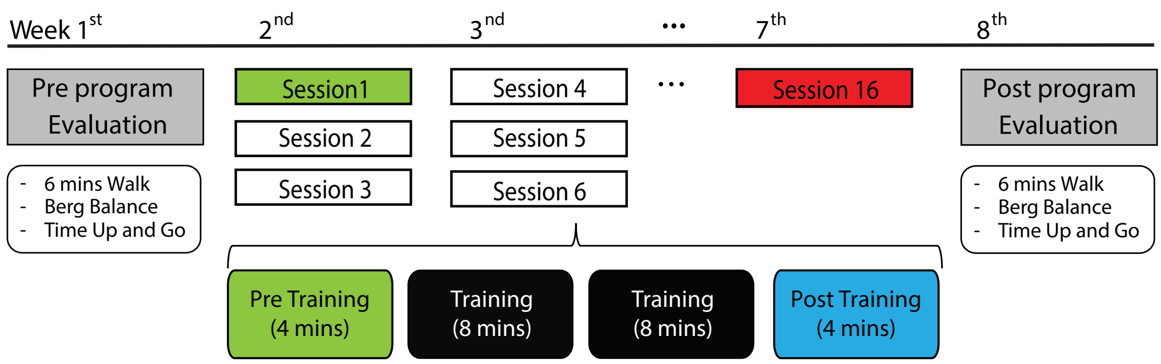


Fig. 3.2 TPAD training protocol. Training consists of 15 training sessions over 5 weeks. Pre-evaluation was conducted 1 week before the first session, and post-evaluation was conducted 2 weeks after the last training session. The 16th session was added after the 15th training session to record free walking on the treadmill with the same speed as session 1.

3.3.3 Data analysis

All data were divided into gait cycles starting with the heel strike on the treadmill. Gait events were detected by the GRFs recorded from the force plate on the treadmill [124]. The markers were attached on the lower limb, pelvis, and C7, as per [113]. Detailed explanation of marker placement is presented in the Supplementary Materials. For pelvic ROM, pelvic center was computed from the geometrical mean of anatomical landmarks (left/right anterior superior iliac and sacrum). Pelvic ROM was calculated by subtracting maximum and minimum displacements of the pelvis within a gait cycle. Step width was defined as the maximum medial-lateral distance between the right and left heel markers during double support period after the heel strike. Step length for one leg was defined as the anterior-posterior distance between the heel markers of two legs at the moment of the leg's heel strike. GRF was filtered with fourth-order low-pass filter with a cutoff frequency of 20 Hz and divided to gait cycle as well.

EMG signals were filtered with fourth-order band-pass Butterworth (40 to 450 Hz), rectified, and then filtered again with fourth-order low-pass Butterworth (6 Hz). For each participant, EMG data of each muscle were normalized to the maximum values recorded during the baseline of the first session. Then, EMG data were divided for each gait cycle and time-normalized from 0 to 100%.

3.3.4 Marker set

Kinematics of the lower limb were recorded using motion capture system (Vicon Bonita, UK). Three dimensional positions of markers were measured at the frame rate of 200 Hz. Twenty four reflective markers were placed on body landmarks in each session of the training. Markers were located on the processus spinosus of the 7th cervical vertebra, median of the most lateral points on the iliac crest, most lateral points of the iliac crests, lateral epicondyle of the femurs, head of fibula, tibialis tuberosity, tips of the lateral malleolus, calcaneus, first metatarsal heads, and tips of the toes. Besides these, markers were placed on the real and lateral parts of the midfemurs. Before every session, a static trial was recorded with seven additional markers on the medial side of the anatomical landmarks. These markers were reconstructed to compute the rotational axes of the joints. Those additional markers were attached on the sacrum, bilaterally on the anterior superior iliac spines, medial epicondyle of the femurs, and tips of the medial malleolus.

Table 3.1 Participant information and clinical measurements. Measurements on participant characteristics and results of clinical test for pre/post TPAD training. The post evaluation is conducted two weeks after 15^{th} training session. Higher score of Berg Balance means better balance with a maximum score of 56.

ID	Age	Sex	Weight (kg)	Height (cm)	FMFCS level	6 Min Walk (m)		Berg Balance scalel		Time Up and Go (s)	
						Before	After	Before	After	Before	After
1	15	M	93	177	II	382	426	49	52	11.98	10.73
2	10	M	42	145	II	317	320	43	42	12.05	11.41
3	9	M	39	140	II	371	418	48	49	7.06	7.87
4	13	M	60	168	II	387	458	53	51	5.73	6.02
5	19	M	87	170	II	334	394	51	51	8.75	8.74
6	17	M	74	168	II	375	406	54	52	9.96	8.92
Mean	13.83	M	65.83	161.33	II	361.00	403.67	49.67	49.50	9.26	8.95
±Std	± 3.92		± 22.69	± 15.04		±28.56	±46.38	±3.98	±3.83	±2.58	±1.95

3.4 Results

In each TPAD training session, participants performed 4 min of free walking before training, 16 min of TPAD walking with continuous downward force equivalent to 10% of body weight, and another 4 min of free walking after training. The free walking data before and after training were collected for comparison to investigate the effects of TPAD in the short term. Participants performed 3 training sessions per week over 5 weeks, with a total of 15 sessions.

The study was designed to observe the training effects on different gait parameters. After finishing all 15 training sessions, an additional session, referred to as the 16th session, was added, in which the children walked with the same treadmill speeds as their first sessions.

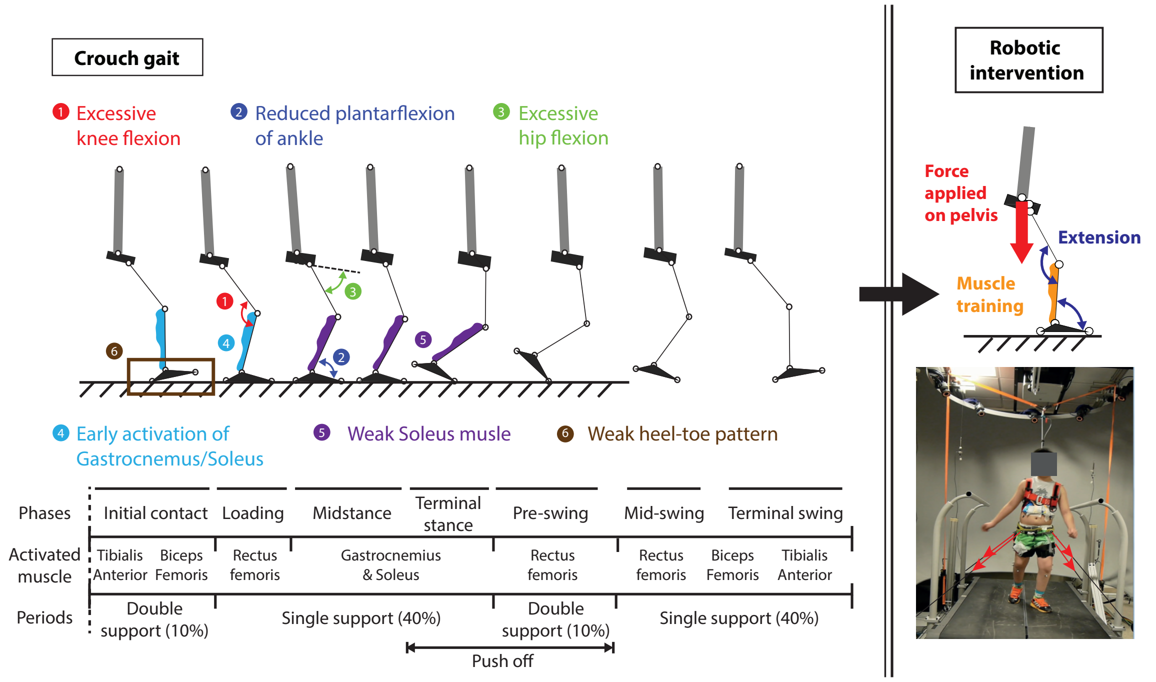


Fig. 3.3 Gait characteristics of crouch gait and TPAD training. Motion capture data from a child with crouch gait during free walking before the training is illustrated, which was collected using a Vicon system. The gait pattern of the crouch gait can be described as excessively flexed hips/knees and reduced plantar flexion of the ankle. TPAD training applies a downward force on the stance leg to improve the muscle patterns necessary for extension of the limbs while bearing the child's weight. Here, the soleus muscle engaged during the single stance phase is intensively trained to improve the crouch gait.

EMG was recorded only during the 1st and 16th sessions. EMGs of soleus and gastrocnemius were monitored for their magnitudes and timing during the gait cycle. The normal EMG pattern of soleus and gastrocnemius in healthy children has a peak around 40 to 50% of the gait cycle after heel strike. However, children with CP often activate these muscles much earlier during the gait cycle, during the initial stance phase. If gastrocnemius muscle activates too early, it hinders the knees from extension because it is a knee flexor [34]. Earlier studies have reported that, when weight is added to the body during walking, soleus and gastrocnemius are active during the latter part of the stance phase. If these muscles are trained to activate during the stance phase of walking with TPAD, we hypothesize that TPAD training will (i) strengthen soleus muscles and (ii) delay the early activation of gastrocnemius/soleus muscles. Especially, TPAD will intensively train the leg during the single stance phase, that is, middle of the stance phase when the lower limb should be straight, whereas only one leg supports the extra downward force. Besides EMG, kinematics and GRFs were also monitored. For the lower limb kinematics, hip, knee, and ankle angles were monitored to verify if (iii) the children extend the joints more during the stance phase. Each joint angle was computed in sagittal plane, as suggested in [119]. As described in Fig. 3.3, children with CP typically do not exhibit distinct heel strike and toe-off during walking, unlike typically developing children [105]. The heel strike plays a role to absorb the shock during the initial contact, and the push-off facilitates forward propulsion of the body [85]. For children with CP, these two gait events are often merged together, and the GRF shows up as a single continuous force, which is undesirable. The distinct pattern of the heel and toe is important to achieve a stable stance and a strong push-off.

3.4.1 Changes in muscle forces and center of mass motion during a single training session: Representative participant

EMGs of two plantar flexor muscles in the lower limbs were measured to observe the effects of the external force on the pelvis. Muscle activations of these two muscles are displayed over the gait cycle, which starts with the heel strike of the leg. The yellow shaded areas in Fig. 3.1(b–c) represent the stance phase of the gait cycle.

Figure 3.1(b) shows the EMG signals for a representative participant in session 1 during free walking and when downward forces were applied. EMG signals are scaled with respect to their maximum values during free walking before training. In Fig. 3.1(b), increased peaks of gastrocnemius and soleus muscles were observed during the stance phase when the TPAD is active. Once the tethers were off, the peaks of gastrocnemius and soleus during post-training were still larger than pre-training. During the mid-stance phase (20 to 40%), the gastrocnemius shows smaller activation during the training and after the training. Figure 3.1(c) shows that the hip and knee are more extended than baseline during the stance phase. Ankle plantar flexion is also increased, closer to a normal plantar flexion angle of -20°, at push-off.

Figure 3.1(d) presents the pelvic translational ROM. Vertical ROM of the pelvis was larger after the TPAD training. The child was able to raise the center of mass more after walking with extra downward force. The second graph in Fig. 3.1(d) shows the trunk angle, which is the angle between the vertical and a line on the trunk, defined by two markers attached to the seventh cervical vertebra and the sacrum. Decreased trunk angle reflects improved crouched posture during post-training compared with pre-training. Vertical GRF was also presented in Fig. 3.1(e). During the training and after training, the pattern of the GRF shows two explicit peaks, characteristic of a distinct heel-to-toe pattern.

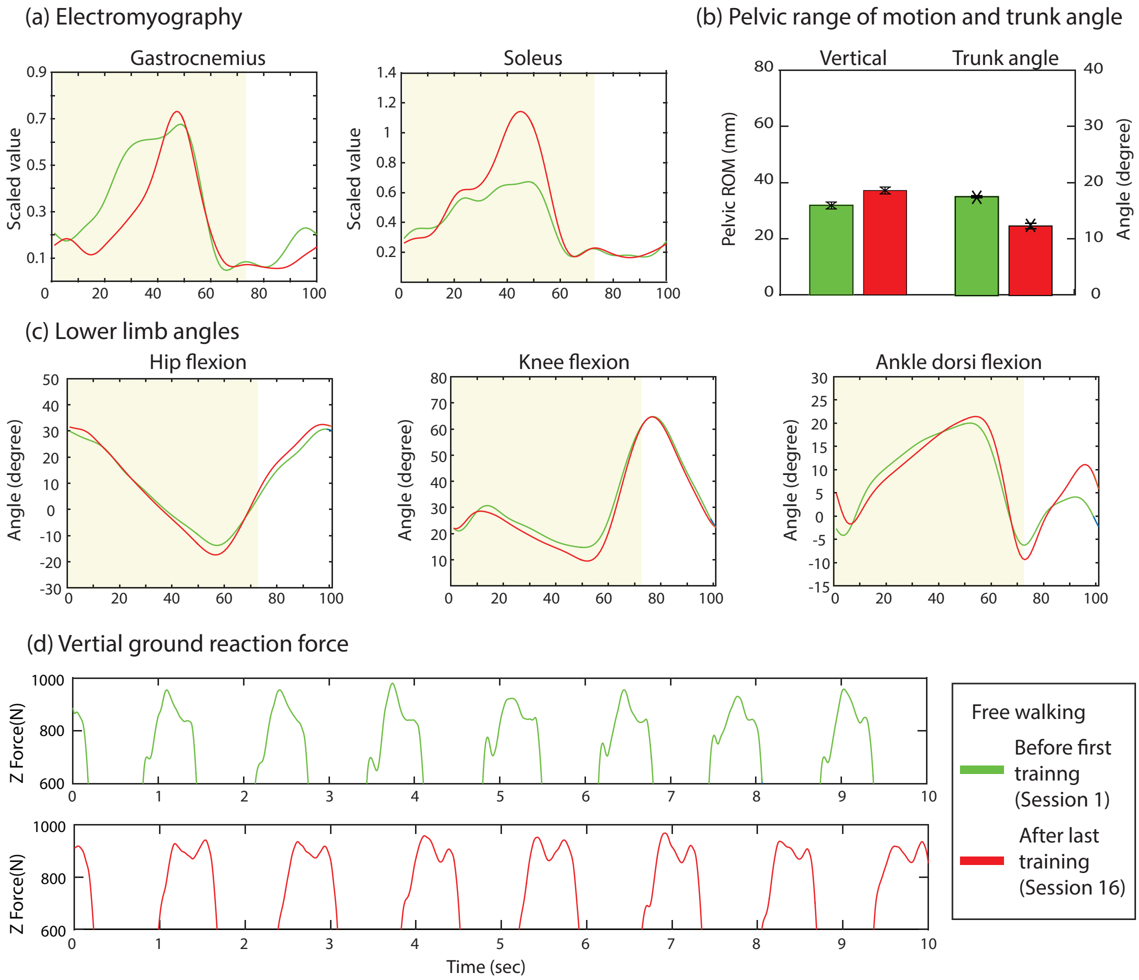


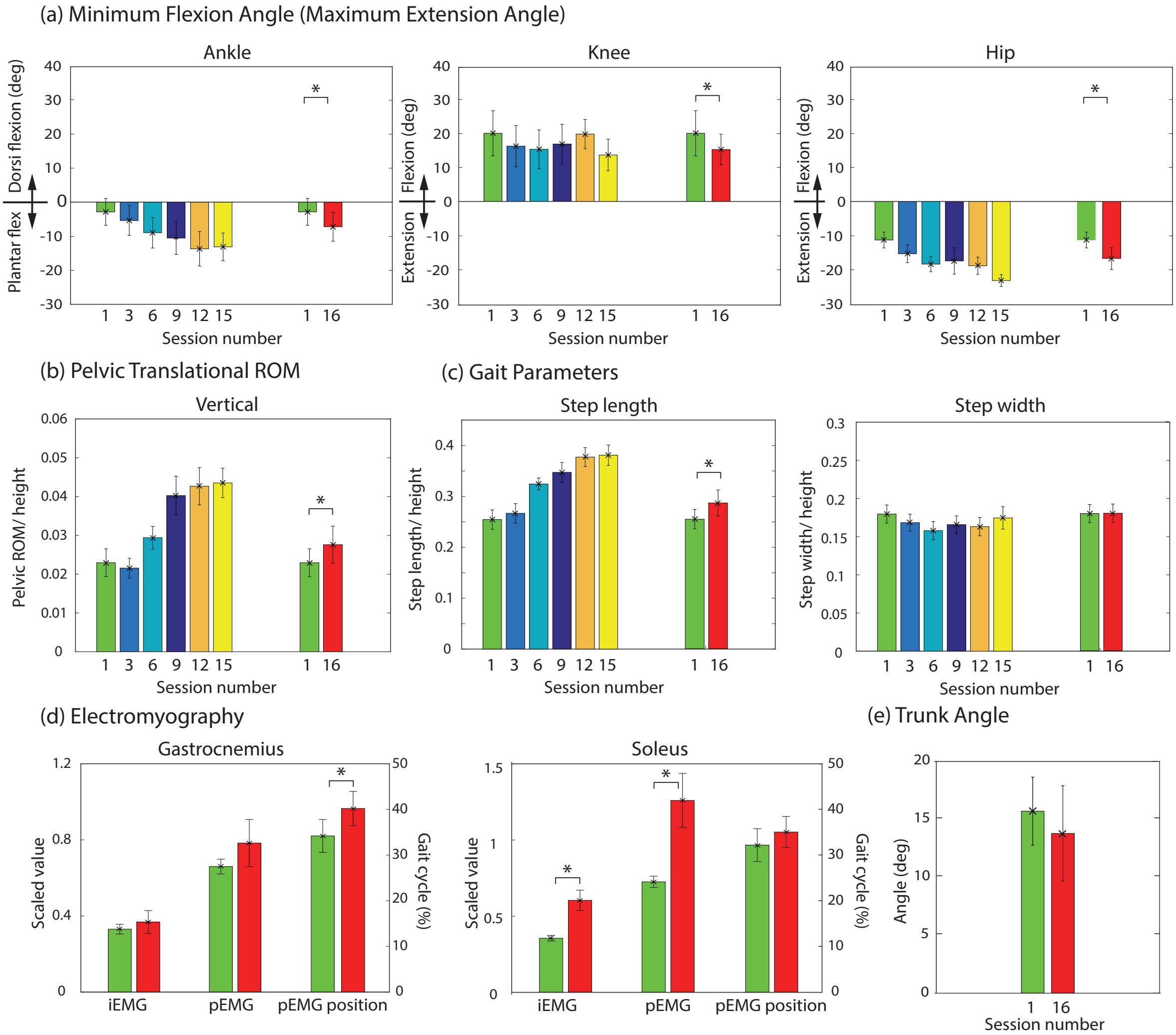
Fig. 3.4 Results of a representative participant with multiple training sessions. Average and SE are presented for the data before and after 15 training sessions during free walking. (a) EMG of muscles. (b) Lower limb angles. (c) Pelvic translation ROM and the trunk angle. (d) Vertical GRFs.

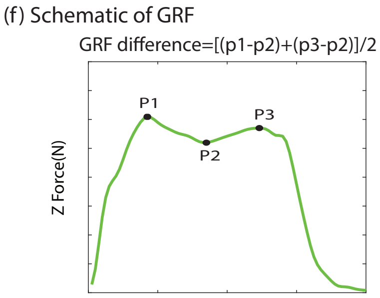
3.4.2 Changes in gait during treadmill walking after multiple training sessions: Representative participant

Figure 3.4 shows different parameters during free walking before (green line) and after 15 training sessions (red line). In these plots, the treadmill speeds before and after 15 training sessions were set the same to remove the effect of walking speed on gait pattern [40]. Figure 3.4(a) shows changes in EMG signals. Both EMG signals were scaled with the maximum value of baseline at the first session (before the first training). The soleus shows higher peaks during stance phase of the gait, and the curve of gastrocnemius is shifted more toward terminal stance phase. These trends were also observed during the training and just after training in session 1; however, the changes are more dominant after the 15th training session. The child showed straighter posture after training. His knee and hip are more extended during the mid-stance phase. Ankle plantar flexion is also increased during push-off with the toes against the ground. The increased ROM of the pelvis in the vertical direction shows that the child was able to push up his center of mass against the gravity. The trunk is straighter after the 15th training. In addition, Fig. 3.4(d) shows two distinct and higher peaks in GRFs, during heel strike and toe-off. This pattern of the GRF was also observed during the first session data, when the tethers were taken off.

3.4.3 Changes in gait during treadmill walking before and after 15 sessions of training: Group results

Six children with diplegic CP participated in the robotic training with TPAD. All of these children presented crouch gait. Participants of ages 20 years and below were recruited and were classified as Gross Motor Functional Classification System (GMFCS) level II. These children could walk independently but experienced difficulty while walking on level or uneven surface. The mean weight of the group is 65.83 ± 22.69 kg, and the mean height is 161.33 ± 15.04 cm. Each child completed the protocol outlined in Materials and Methods.





(g) Measurements of walking ability

Parameter	Before	After	p-value
Overground speed	6.26±0.26	7.27±0.39	0.031
Treadmill speed	3.70±1.08	7.12±1.03	0.028
Below data are collected with same treadmill speed.			
Ankle ROM (deg)	28.24±7.84	31.96±6.90	0.046
Knee ROM (deg)	36.32±8.48	39.57±11.32	0.173
Hip ROM (deg)	41.12±8.37	44.46±12.05	0.116
Symmetry (%)	7.8±3.01	5.6±3.77	0.046
Toe clearance	0.048±0.012	0.058±0.015	0.028
GRF difference	2.30±1.01	4.66±1.27	0.031

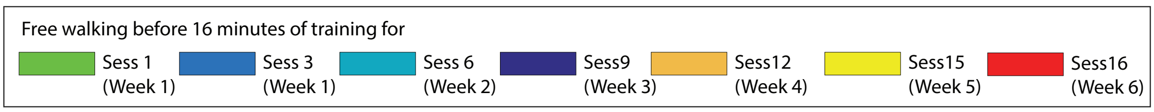


Fig. 3.5 Group data during free walking between sessions 1 and 16. For (a) to (c), the left side of the graphs presents data for each week (sessions 1, 3, 6, 9, 12, and 15), and the right side presents data for sessions 1 and 16, which have the same walking speed. From sessions 1 to 15, the treadmill speed was increased every first session of each week, as described in Materials and Methods. (d) EMG is scaled with the maximum value of baseline in session 1. (e) Trunk angle during walking. (f) The definition of the GRF difference is illustrated, and its group value is presented in (g). Mean and SEs are presented for six participants (n = 6), except EMG (n = 5).

Group data of all six children are presented in Fig. 3.5. In Fig. 3.5(a-c), left graphs show free walking over different training sessions, and right graphs show free walking for sessions 1 and 16 with the same walking speed. Nonparametric statistical analysis was conducted only for sessions 1 and 16. Wilcoxon signed-rank test was used with = 0.05 [123], because the data did not satisfy the normal distribution due to small number of participants. All lower limb–related parameters are averaged for left and right legs. Figure 3.5(a) demonstrates the minimum flexion angle, namely, maximum extension (or maximum plantar flexion), for hip, knee, and ankle joints. These parameters show whether the participants improved extension after being trained with TPAD. All hip, knee, and ankle angles showed decreasing trend toward smaller joint angles. The ankle angle showed a significant decrease before and after 15 sessions of training: for ankle, $P = 0.028$ with the effect size of $r = 0.635$; knee, $P = 0.046$ with $r = 0.575$; and hip, $P = 0.028$ with $r = 0.635$. Because the effect size is larger than 0.5, the intervention has a large effect in the measured variables [123]). Pelvic translational ROM increased over training in the vertical direction as well ($P = 0.046$ with $r = 0.575$). Step length and step width were also compared before and after training. Step length increased over the training and showed a significant change between sessions 1 and 16 ($P = 0.046$ with $r = 0.575$), whereas step width did not change over training and after 15 sessions of training. Pelvic translational ROM and gait parameters were scaled with the height of the participants. Trunk angle showed a decreasing trend in Fig. 3.5(e), but no significance was found. All these parameters' data during the training with the downward force are included in Fig. 3.6.

Walking ability of children is often evaluated by walking speed, step length, and foot clearance. Significant increases were observed in both overground and treadmill walking speed of children after the training. Overground walking speed was computed from the Six Minute Walk test. In addition, treadmill walking after 15 sessions of training showed increased step length, symmetric gait, and better foot clearance, as shown in Fig. 3.5(g). Symmetry of each child was computed by comparing the normalized stance time between right and left legs. According to Eq. (3.1) 1, the value is close to 0, when the child presents a

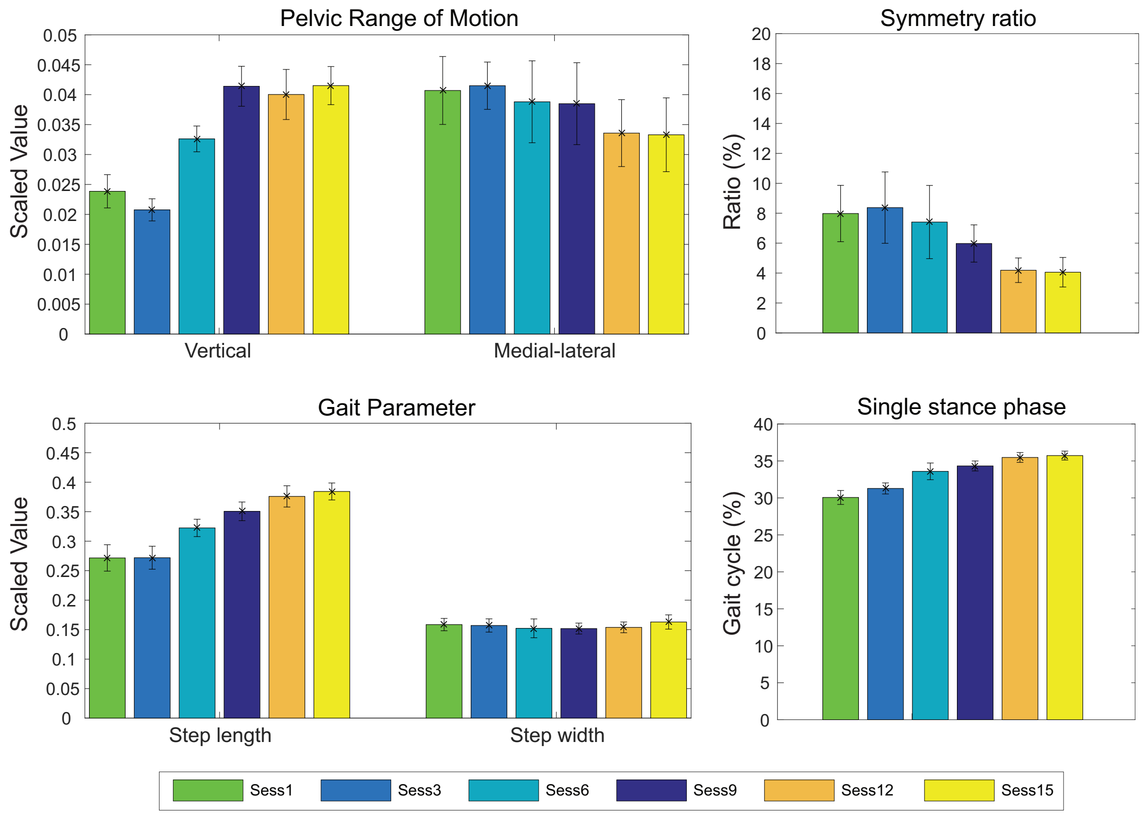


Fig. 3.6 Pelvic ROM and gait parameters during the training. Treadmill walking was recorded with the extra downward force for session 1, 3, 6, 9, 12, and 15. For the first training session of each week, the speed of treadmill is increased, if the child was able to walk with faster speed comfortably. The data were recorded at the 7th minute of each training session. Pelvic range of motion and gait parameters are scaled by height of each individual. Mean and standard error are presented.

symmetric gait.

$$\text{Symmetry}(\%) = \left| 1 - \frac{\text{right stance time}}{\text{left stance time}} \right| \times 100 \tag{3.1}$$

EMG of gastrocnemius and soleus is presented in Fig. 3.5(d) for the group. We investigated the magnitudes and patterns of the EMG activity in two extensor muscles for ankle, gastrocnemius and soleus, before and after 15 sessions of training. EMG values are filtered, rectified, and then scaled with the maximum value of the baseline in session 1. If the muscles in the 16th session are firing stronger than the first session, the scaled value can be above 1. After all post-processing (see Materials and Methods), three different parameters were computed to extract the characteristics of the EMG pattern:

1) iEMG: mean value of post-processed EMG over the entire gait cycle.

2) pEMG: peak value of post-processed EMG of each gait cycle.

3) pEMG position: position of the peak value in terms of the percentage of the gait cycle.

If the pEMG is larger than iEMG, not only the EMG is increased during the entire phase but also the shape is changed toward a sharper cone, the normal EMG pattern of soleus and gastrocnemius.

Figure 3.5(d) shows the magnitudes of integrated and averaged EMG (iEMG) and peak EMG (pEMG), with higher magnitudes after 15 training sessions. Normal EMG patterns of these muscles have sharp peaks around 40 to 50% of the gait cycle. The gastrocnemius peak after training has shifted to the right, closer to 40% of the gait cycle ($P = 0.043$ with $r = 0.640$), and the soleus significantly increased the magnitude for both iEMG and pEMG (both $P = 0.043$ with $r = 0.640$).

Another important observation is that children develop distinct heel strike and toe-off as a result of this training. Heel-to-toe pattern can be characterized by peaks and valleys of the GRFs [5]. Statistical analysis was conducted for sessions 1 and 16 with the same walking speed. We define "GRF difference" to characterize the heel-to-toe pattern of walking using the equation below:

$$\text{GRF difference} = \frac{[\text{first peak(P1)} + \text{second peak(P3)}]}{2} - \text{valley(P2)}. \tag{3.2}$$

The first peak (P1) in Fig. 3.5(f) indicates the force during heel strike, and the second peak (P3) is the force during push-off of the stance foot. GRF difference was scaled for each child's weight. GRF difference values for sessions 1 and 16 are presented with the P value in Fig. 3.5(g). The training promoted a heel-to-toe pattern that stabilizes the stance leg during initial contact and propels the body forward. Clinical evaluations were conducted twice, 1 week before the first training and 2 weeks after the last training. Six Minute Walk test (before:

361 ± 28.56 m; after: 403.67 ± 46.38 m), Pediatric Berg Balance Scale (BBS) (before: 49.67 ± 3.98; after 49.50 ± 3.83), and Timed Up and Go (TUG) test (before: 9.26 ± 2.58; after: 8.97 ± 1.95) were evaluated, but no significant improvement was found after the training for BBS ($P = 0.785$ with $r = 0.079$) and TUG test ($P = 0.345$ with $r = 0.272$). Six Minute Walk test showed significant improvement ($P = 0.028$ with $r = 0.635$). Individual values are presented in table 3.1.

3.5 Supplemental results

3.5.1 Ground reaction force data of all six children

Ground reaction force for both right and left leg of six children were measured using force plate on the treadmill. Figure 3.7 presents vertical ground reaction force during free walking averaged over 8 gait cycles. Subjects with flat foot walking showed a definite heel strike at the end of the training (Session 15). Session 16 was added to evaluate walking with the same walking speed. The data of session 1 15 are also presented for both free walking and during the training, as shown in Fig. 3.8. The walking speed was chosen for the first training session of each week, based on the walking ability of children. During the training, subjects demonstrate larger vertical force than the free walking data.

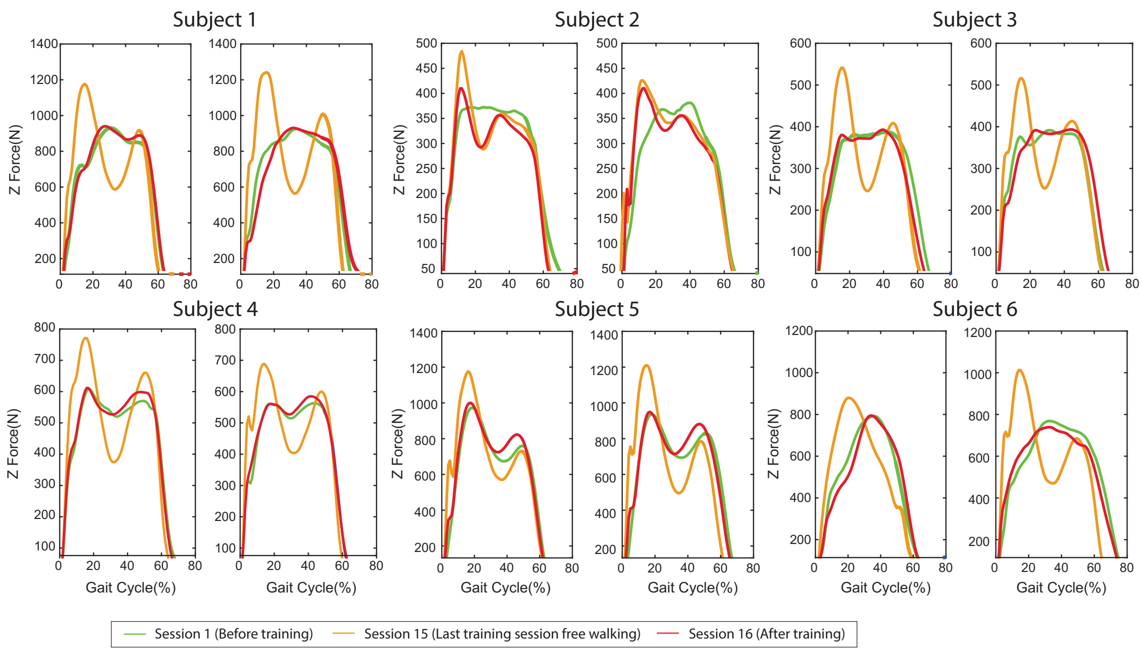


Fig. 3.7 Vertical GRF of sessions 1, 15, and 16 for all six children during free walking. Ground reaction forces of both feet are shown for each child. Session 1 and 16 were conducted with the same walking speed.

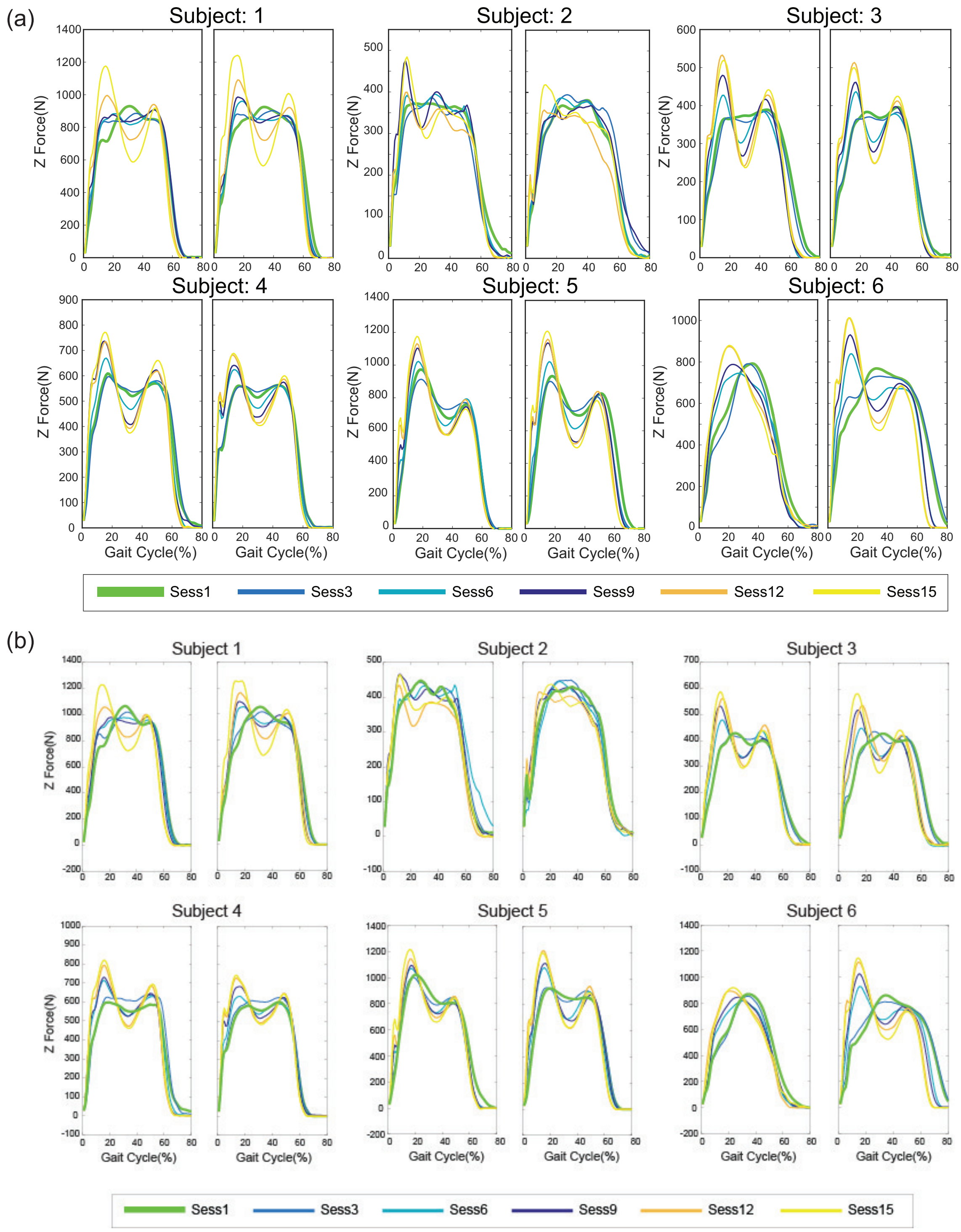


Fig. 3.8 Vertical GRF of all six children during free walking and training with downward force. Ground reaction force of session 1, 3, 6, 9, 12, and 15 are presented for right foot and left foot. Children initially had a nearly flat foot gait which shifted to heel-to-toe pattern of walking after 15 sessions of training with TPAD. (a) Free walking on the treadmill and (b) walking with extra downward force was recorded. For the training data, the data was recorded at the 7th minute of each training session.

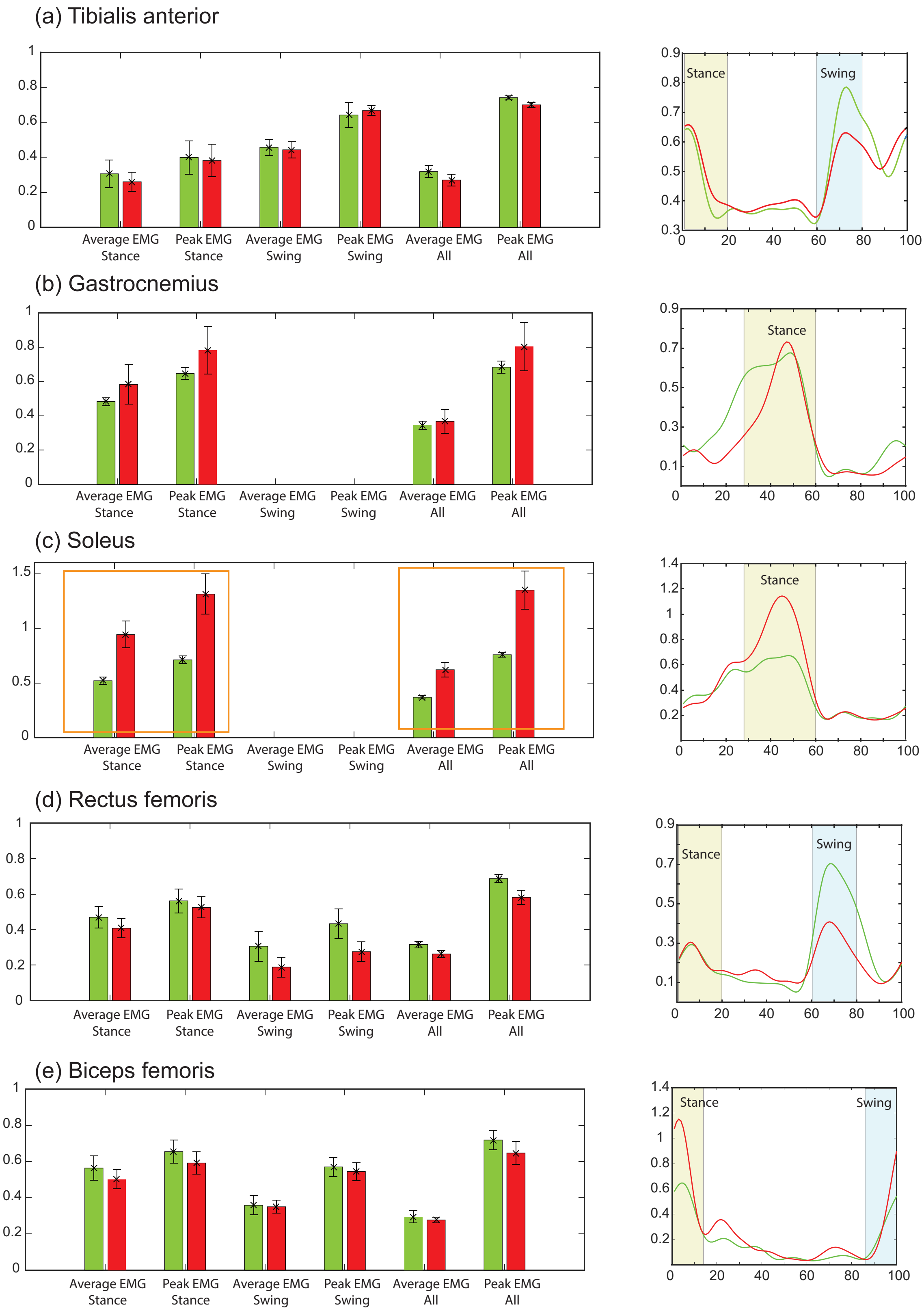


Fig. 3.9 EMG for five different muscles before and after 15 training sessions. Stance phase was highlighted in yellow and swing phase was in blue shade.

3.5.2 Gait and EMG parameters

In Fig. 3.5, the data during the free walking through session 1, 3, 6, 9, 12, and 15 are presented. The corresponding data during the training with the extra downward force are shown in Fig. 3.6. Pelvic translational ROM in medial-lateral and vertical direction, step length, step width, and single support phase (SS) are demonstrated. Symmetry ratio is computed in the same manner with Eq. (3.1).

The trend that was observed during free walking is similar to the trend during the training with TPAD. Increased pelvic ROM in vertical direction, increased step length, better symmetry, and longer single stance were observed during the training as well. These two data sets demonstrate that the training with downward force induced the positive changes in the gait of cerebral palsy children during free walking.

3.5.3 Walking speed of children with crouch gait

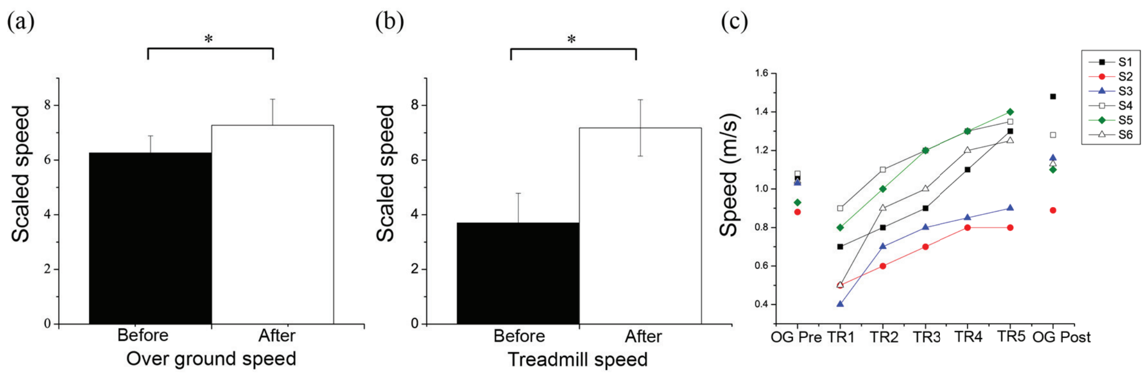


Fig. 3.10 Walking speed of children with cerebral palsy. (a) Over ground walking speed before and after participating in the training. (b) Walking speed on a treadmill before and after participating the training. Speed in (a) and (b) are scaled by height of the individual child. (c) Walking speed without scaling for overground walking (OG) and treadmill walking (TR) during 1st, 2nd, 3rd, 4th, and 5th week. Six CP children who participated in the TPAD training increased both over ground and treadmill walking speed with statistical significant level $\alpha = 0.05$.

3.5.4 Subject information and results of clinical outcomes

Table 3.2 Spasticity of ankle and knee joints.

ID	MAS Ankle				MAS Knee			
	Pre		Post		Pre		Post	
Right/Left	Right	Left	Right	Left	Right	Left	Right	Left
2	3	3	3	3	2	2	2	2
3	2	2	2	3	$1+(1.5)$	1+	1+	2
4	2	2	$1+(1.5)$	2	2	2	1+	2
5	1	1	1	1	1	1	2	2
6	3	3	2	2	1+	1+	2	2
Average	2.2	2.2	1.9	2.2	1.6	1.6	1.8	2

* Spasticity is not evaluated in the subject 01.

Spasticity was measured as a reference, if the training worsen the spasticity. However, there wasn't any significant change between before and after the training.

Table 3.3 Response of the subject and parents from clinical chart.

ID	Comments from the doctor or subject
1	Patient 1 is interested in continuing the therapies with the walking study. He wishes it did not end. Patient 1 reports his whole walking changed. He says he feels more comfortable with walking now and has better balance.
2	Mom reports he is improving. He can step up onto the curb by himself without holding hands. Mom reports his left foot is better.
3	Completed TPAD a month ago, therapist at school says he no longer requires a walker for ambulating and would like to transition him to using bilateral forearm crutches. Mom reports that his muscles are better after the treadmill and that he is walking with a heel strike now. Gait pattern has improved and he doesn't not fall anymore.
4	Before the training, the child complaint "I feel like my les will never get better" and felt depressed with the way he walks. He completed TPAD study 2 months ago and he and his mother report that he had great outcomes from study. His gait improved and he was happy with this change.
5	He is s/p TPAD study and had good outcomes following this. He had decreased crouching in his gait improved endurance, and improved bike riding. As he had great outcome following participation in TPAD study and continues to have functional gait with good balance and decreased crouching. He does not require repeat botox injections at this time.
6	He got taller during the gait and walking speed was increased after TPAD training.

3.6 Discussion

This study shows the feasibility that training with TPAD using downward pull force can improve walking of CP children with crouch gait. The lower limbs have increased extension after the training, showing stronger soleus muscles. Improved vertical pelvic ROM and decreasing trend of trunk angle are also observed. In addition, children who participated in this study showed improvements in walking ability, for example, step length, walking speed, symmetry, and foot clearance. Currently, there is no well-established physical therapy or strengthening exercise for the treatment of crouch gait [51]; TPAD training with downward pelvic pull could be a promising intervention for these children.

After training, children presented increased extension of the knee and plantar flexion of the ankle. This is explained by the increased activity of the soleus muscle, which prevents the progression of the tibia during the stance and allows the knees to extend due to momentum of the center of mass. The improved timing of the gastrocnemius also contributed to knee extension, because the gastrocnemius activated less during the early stance phase to prevent knee flexion of the crouch gait. Improved muscle activation of soleus and gastrocnemius also increased the plantar flexion of the ankle during the push-off phase. However, hip extension cannot be directly explained by those two muscles. Hip extension can increase due to activation of the muscles that were not measured in the present study, such as gluteus maximus, adductor magnus, semimembranosus, and semitendinosus. Another possible explanation is that the changed posture of the knee and ankle enabled the hip to extend more easily. In crouch gait, the hip requires larger torque for extension because the knee is leaning forward, and the body weight vector creates a larger moment arm to flex the hip [85].

In the past, studies were conducted to improve crouch gait of children with CP by strengthening hip and knee extensors using bench press or elastic rubber bands [23, 107, 32]. These studies targeted knee extension and reported inconsistent outcomes after strength training [102]. Instead of focusing only on the knees, TPAD chooses to train all leg extensor muscles, especially the soleus. Strength training of extensor muscles during walking with

TPAD might be more effective than conventional strengthening therapy for two reasons. First, loading is an important sensory cue that facilitates activation of the extensor muscles during gait [36, 1, 11]. With the additional downward force on the pelvis using TPAD, this sensory cue is emphasized to induce activation of the appropriate motor pool for load bearing. Second, the activation is phase-dependent when the limbs support weight [36, 11]. This may explain why training with bench press or stretching with rubber bands does not transfer to improvements in gait [96].

Strengthened ankle extensor muscles reinforce the push-off force of the lower limbs. As the leg swings further with increased step length and toe clearance, children attain a better heel strike. Stronger heel strike was also reported with treadmill training on an inclined surface [62, 115]. Although this training improved heel strike of children with CP, the underlying mechanism in this study is different from the current study. Walking on an inclined surface strengthens weak tibialis muscles to increase dorsiflexion of the foot and lift the toe for heel strike. Another benefit of the strong push-off is that walking speed becomes faster with increased step length. Similar results have been reported in conventional treadmill studies with or without the partial weight support [12, 27, 21, 39]. In these studies, a physical therapist assisted the child with manual corrections and verbal instructions, which led to improvements in walking [21]. Instead of a physical therapist assisting in walking straight, TPAD can provide a robotic environment for the children to strengthen extensor muscles compared with ordinary walking. This device can potentially reduce the effort and labor of the physical therapists so that they can focus on higher-level management, dosage, and monitoring of the patients.

The BBS and TUG tests were introduced as secondary measures of this study to determine how the training results translate to other tasks. For TUG test, we found that only half of children improved. This might be because children who did not show improvement already had a higher performance on this measure before training. Their TUG values were between the values of typically developing children and CP children with GMFCS level I [79].

Although these children had high performance in TUG, their posture was different from the typically developing children. Unfortunately, TUG score only measures the total speed and does not capture the quality of each task. TUG is measured from a dynamic task consisting of different movements; however, BBS mostly consists of static tasks [117, 52]. This might be the reason why BBS training did not improve after TPAD training. A study over 5 weeks may be insufficient to transfer the training effects to other tasks that are not directly related to the trained task. Among these three clinical scales, we should note that the Six Minute Walk test showed improvements. Six Minute Walk test is a self-paced, submaximal test that assesses functional capacity for walking over a prolonged distance [109, 106]. It reflects exercise tolerance and endurance required for the performance of active daily living [106]. Here, we asked children to walk within a submaximal environment, by augmenting weight, for longer than 6 min. We think that the training effect of the present study is better estimated by the Six Minute Walk test that is more analogous to the task in the TPAD training, rather than BBS or TUG tests.

Future studies warrant some attention. When recruiting children for TPAD training, independent walkers with GMFCS level I or II are recommended. TPAD training provides a downward force on the pelvis, and any hand holding of the rails will diminish the force delivered to the joints of lower limbs and reduce the training effects. For children who are more severely affected, other robotic interventions with assistance or guidance force might be more desirable [108]. Second, the present study was designed to compare parameters before and after the training with the same treadmill speed (Fig. 3.5). The training improvements, as stated here, are conservative because the treadmill speed was kept the same before and after training for scientific comparison. Once trained, the optimal walking speed of the children was higher, and their gait should be measured and documented at that speed. Future design of the experiment should include overground walking data for evaluation of the training. Third, this study does not include a randomized control group. The impact of natural ongoing development of the children should be considered within the result. However, there are studies that show that there are no significant changes due to the natural development of CP children within 4- to 6-week period, unless they undergo an intervention [95, 19, 56]. These studies support this idea by measuring GMFM D (standing function) and GMFM E (walking function) in children with CP. As future study, a larger number of participants with a control group should be organized. Last, this pilot study applied 10% of downward force as a start of this new training paradigm. Different dosage in terms of the amount of the weight and period should be tried and compared to maximize the benefits of this training. Large population of participants with CP will be recruited in the future with different GMFCS levels to increase the power of this study. We are also considering studies targeted at children with hemiplegic/quadriplegic CP.

Feedback from parents and children involved in this study was consistent: They reported improved posture, stronger legs, and faster walking speeds. Our results show that intensive and well-designed paradigms of gait training with robotic devices, such as TPAD, can yield improvements in gait, as has been postulated in previous studies with CP children [39, 116]. Our study confirms these observations, as EMG magnitudes and the pattern of muscles improved after 15 sessions of gait training. Future works should investigate more stable and permanent functional recovery through longer interventions.

3.7 Conclusion

The present study showed the feasibility to use this training method to strengthen lower limbs and correct the posture of children with crouch gait. Downward force increased the activation of the soleus muscle and delayed the early activation of the gastrocnemius muscle. This change of muscle pattern enabled straighter lower limb during the middle of stance phase and improved walking. Because there are no well-established physical therapies or strengthening methods for children with crouch gait, our training method may be a promising intervention for these children. This new training could be used solely as a strength training device in the clinic or combined with conventional treatment such as Botox or ethanol injection to reinforce both methodologies.

Chapter 4

Simulating hemiparetic gait and asymmetric weight bearing in healthy subjects using TPAD with a novel controller

In both Chapters 2 and 3, tethered pelvic assist device (TPAD) provided interventions that were bilaterally applied on both legs. Chapter 4 introduces a training strategy that can apply an asymmetric force field which acts differently on each leg. The asymmetric force field is created by a closed loop controller that was motivated by the results in Chapter 2. This training is expected to be used for patients who have impairments on only one leg and results in asymmetric weight-bearing while walking.

4.1 Asymmetric gait of hemiparesis patients

Hemiparetic gait is a common walking abnormality seen in patients with stroke and cerebral palsy [67]. Annually, approximately 800,000 people have an incidence of stroke [35] and 1 out of 1000 children are born with hemiplegic cerebral palsy [120]. Impairments in these patients are mainly on one side of their body which results in asymmetry in gait. Their step lengths with the two legs and the step widths on the two sides with respect to the vertical projection of the body center of mass are also different. They tend to have a longer stance phase on the non-paretic limb. Besides temporal or spatial gait parameters, asymmetric weight bearing is also a major problem for these patients because of two major reasons: (i) If the patient relies excessively on the non-paretic limb during walking [55], the subject will overload the joints and suffer from pain and fatigue. (ii) As paretic limb will be less used during walking, the muscles of this limb will be negatively affected due to disuse [84]. A recent observational video study performed in a long-term care facility showed that incorrect weight shift and the resulting kinematics was one of the main causes for falls in these patients [91]. Another study demonstrated that gait asymmetry became worse after the incidence of a stroke [83].

In the past, researchers have focused on improving gait symmetry using different methods in physical therapy. Previous studies have showed that the current training methods with physical therapy do not improve gait symmetry [37, 38]. This may be due to the nature of the physical therapy which is designed to change the muscle tone while sitting or standing. [37, 38]. Some physiotherapy programs attempt to achieve symmetric weight bearing using external weights. The external weights are added on either the affected or unaffected limb to alter the mass properties of the two legs with the goal to promote symmetric weight bearing [59, 90]. Another method that was suggested to change the kinematic characteristics of the non-paretic limb. Shoe wedges were designed and added asymmetrically to promote longer stance phase and weight bearing of the paretic limb [99]. Body-weight supported treadmill training was also reported to have similar effects on gait symmetry during overground training [74].

Besides physical therapeutic methods, various robotic gait trainers have been developed to improve walking ability of hemiparetic patients. A robotic leg exoskeleton was reported to improve the symmetry of the temporal parameter of the paretic limb [44]. A soft exosuit

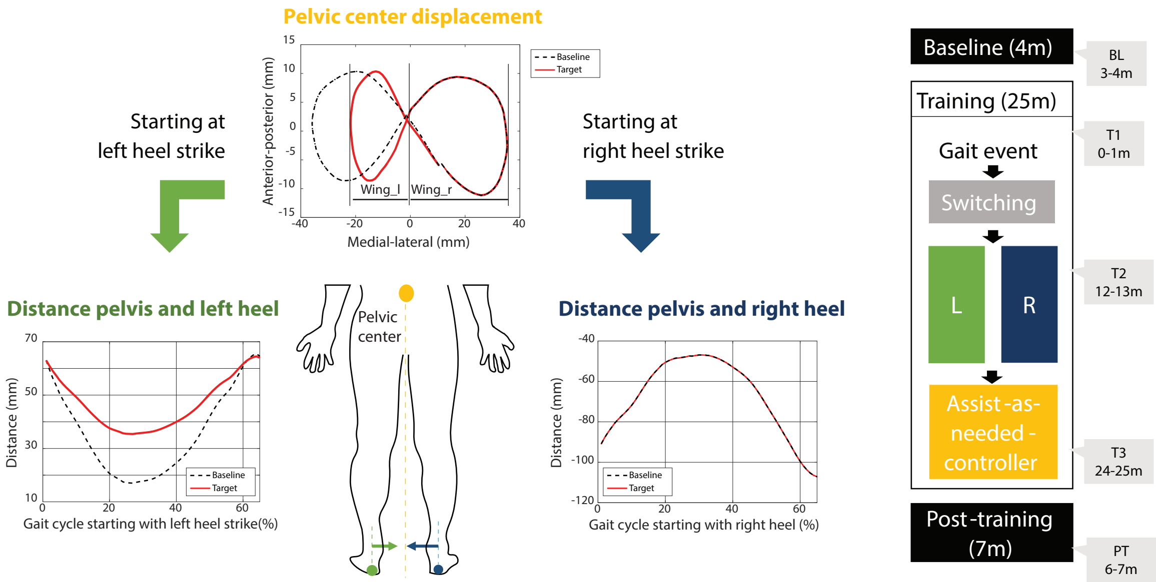


Fig. 4.1 (a) Design of the target pelvic trajectory to generate asymmetric weight bearing in healthy subjects. (b) Training protocol for baseline, training, and post-training sessions is presented.

on the ankle presented the improvement in the forward propulsion asymmetry of stroke patient [9]. In addition, a pelvic device was designed to improve the asymmetry of pelvic obliquity. The results showed the feasibility of modifying the pelvic obliquity of healthy subjects during walking [88]. More recently, a cable-actuated device that assisted in shifting the weight between both legs was tested for children with cerebral palsy [121].

4.2 Methods

In this study, assist-as-needed control method is used to simulate asymmetric gait of hemiparesis patients. This controller provides a lateral force on the pelvis in order to change the symmetry in weight bearing on legs. The medial-lateral directed force is computed to be proportional to the deviation in the trajectory of the pelvic center from a target pelvic trajectory, as shown in Fig. 4.1(a) [49]. As the center of the pelvis lies close to the center of mass of the human body [114], we hypothesized that such a force field controller that uses the deviation of the pelvic trajectory from a target trajectory could provide appropriate forces to promote changes in weight bearing. This force field controller is a closed-loop controller that responds to current motion of the pelvis.

This controller is implemented using a cable-actuated device called the Tethered Pelvic Assist Device (TPAD) [49]. The performance of the controller is validated by an experiment conducted on eight healthy subjects. In this experiment with healthy subjects, our goal was to promote a higher weight bearing on the right leg. Ground reaction forces (GRF) in the lateral and vertical directions were collected along with the kinematic data to understand the adaptation of the human gait with the guidance force at the pelvic center. While this experiment was conducted to make healthy subjects walk asymmetrically under the effect of the controller, in the future, this controller will be tested on patients with hemiparetic gait to promote symmetric walking. Details of the assist-as-needed control method is described in Section 2.1.

4.2.1 Human Experiment

Eight healthy adults signed a written consent form approved by the Institutional Review Board (IRB) of Columbia University before the experiment. Average height was 171.4 ± 4.5 cm, average age was 61.5 ± 7.4 kg, and average age was 29 ± 2.5. All subjects walked with their preferred treadmill speed (average 0.93 ± 0.05 m/s). Markers were attached on the right & iliac crest, sacrum, heels, and toes of the subject.

4.2.2 Protocol

The target trajectory was created from the recorded baseline data of each subject in order to induce asymmetry in the gait. Baseline data was high pass filtered with a cut off frequency of 0.4 Hz to remove slow translation on the treadmill platform. The filtered data was divided by gait cycles. The left side of pelvic trajectory was reduced by 40 % compared to baseline as shown in Fig. 4.1. This percentage was chosen to provide abnormal gait out of the standard deviation of normal gait.

The experimental protocol consisted of three sessions, as seen in Figure 4.1(b). During the baseline session (BL), subjects were asked to walk for four minutes and the normal walking pattern was recorded. Data was collected in the last minute of this session. The training session (T) was for twenty five minutes and the assist-as-needed force was applied on the subject with the TPAD. Data was collected every eight minutes and were denoted here as T1, T2, and T3. For the post-training session (PT), all cables were removed and subjects walked for another seven minutes. Data was collected at the last minute of post-training and is referred to as PT. Subjects were asked to place each foot on the corresponding treadmill belt to measure GRFs separately for right and left foot.

4.2.3 Data analysis

All data was recorded for one minute and divided into gait cycles. A Gait cycle starts with the right heel strike on the treadmill and was calculated from the foot markers and the sacrum marker [124]. The divided data was time normalized to 100 % gait cycle. To validate the performance of the subjects and understand how they achieved the target trajectory, various parameters (P) were estimated with a Symmetry Index (SI) which is defined as [53, 18]:

$$SI(\%) = \frac{(P_{right} - P_{left})}{(P_{right} + P_{left})} \times 100. \quad (4.1)$$

First, activations of muscles that participate in the weight bearing are measured. Electromyography (EMG) signals were filtered with fourth order band-pass Butterworth 40–450 Hz, rectified, and then filtered again with fourth order low-pass Butterworth, 6Hz. For each subject, EMG data of each muscle was normalized with respect to the maximum values recorded during the baseline session. Second, vertical and lateral GRFs were compared between right and left foot to evaluate the symmetry of the weight bearing. It was filtered with fourth order low pass filter with cutoff frequency 20 Hz and divided into gait cycle as well. Both EMG and GRF data were integrated and averaged over the gait cycle. In addition, lateral leg angle was computed by the angle from the vertical line (black dash-dotted line in

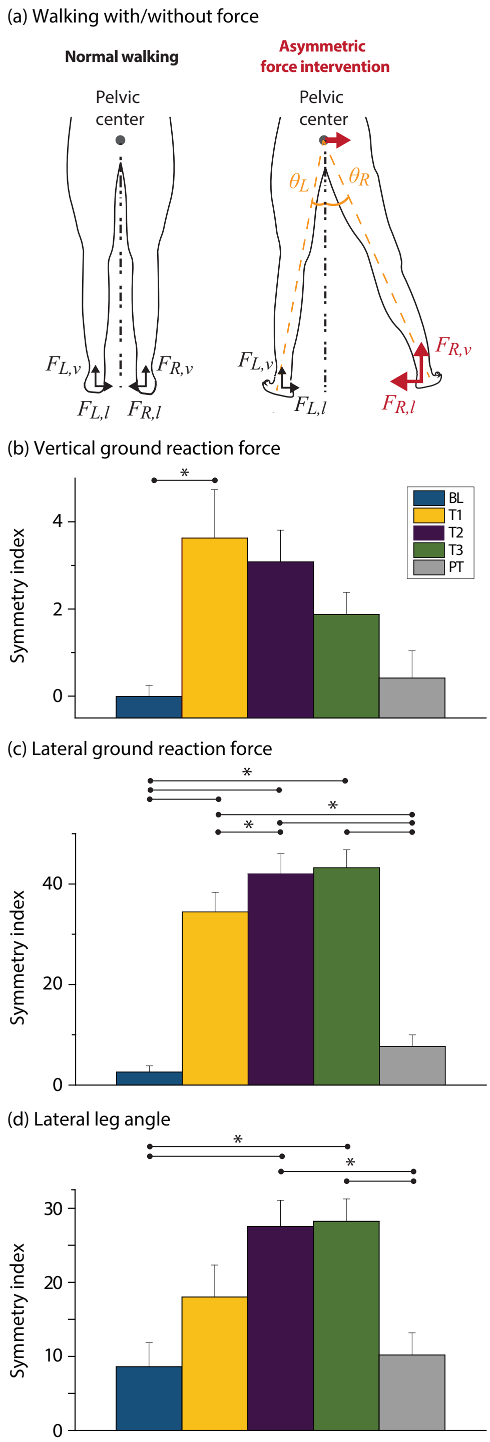


Fig. 4.2 (a) Schematic of free body diagram of a subject without and with the asymmetric force intervention, where $F_{R,v}$ and $F_{L,v}$ are the vertical ground reaction force (GRF) while $F_{R,l}$ and $F_{L,l}$ are the lateral GRF on the right and left foot respectively. θ_R and θ_L are the lateral leg angles of the right and left foot respectively. Under the asymmetric force intervention on the pelvic center, the parameters of $F_{R,v}$, $F_{R,l}$, and θ_R on the right side increased relative to the left side. Such asymmetric increases are estimated using symmetry index for (b) vertical ground reaction force (GRF), (c) lateral GRF (n=6), and (d) lateral leg angle with $p < 0.05$.

the Fig. 4.2(a)) to the inclined line (yellow dashed line in the Fig. 4.2(a)). This inclined line is drawn from the sacrum to the median of first and fifth metatarsal for each leg, denoted by θ_R (right) and θ_L(left).

Also, gait parameters were computed for each session. Step width was defined as the maximum medial-lateral distance between the right and left heel markers during the double support period after the heel strike. Step length for a leg was defined as the anterior-posterior distance between the heel markers of two legs at the moment of the leg's heel strike. These values were normalized by each subject‘s height. Stance time for each foot was computed from heel strike to toe off and scaled with the period of each gait cycle.

For each parameter, one minute's data was averaged to compute the representative value. Repeated ANOVA was run among baseline, training, and post-training sessions ($\alpha = 0.05$). In advance, Mauchly's test was conducted to check the assumption of sphericity. If sphericity was violated then the Greenhouse-Geisser correction was performed for repeated ANOVA. For the pairwise comparison, Bonferroni-Holmn correction was used.

Table 4.1 Gait Parameters of stance phase, step width, and step length.

Parameters		BL	T1	T2	T3	PT
Stance (%)	Sym. Index*	-0.087 ±0.239	1.347 ±0.646	0.680 ±0.353	0.123 ±0.255	0.166 ±0.381
	Right	66.359 ±0.260	66.628 ±0.655	66.226 ±0.348	65.958 ±0.302	65.503 ±0.762
	Left	66.474 ±0.231	64.839 ±0.279	65.329 ±0.281	65.797 ±0.315	65.306 ±0.968
Step Width	Sym. Index	0.530 ±0.440	-1.177 ±0.329	-0.511 ±0.778	-0.744 ±0.578	1.735 ±1.066
	Right*	0.135 ±0.002	0.185 ±0.005	0.172 ±0.007	0.172 ±0.008	0.156 ±0.014
	Left*	0.133 ±0.002	0.189 ±0.005	0.176 ±0.009	0.175 ±0.009	0.153 ±0.015
Step Length	Sym. Index	-0.787 ±0.439	-3.376 ±1.413	-4.209 ±1.677	-1.695 ±0.947	-2.472 ±2.123
	Right	0.309 ±0.006	0.278 ±0.004	0.270 ±0.013	0.295 ±0.005	0.292 ±0.014
	Left	0.313 ±0.005	0.298 ±0.008	0.292 ±0.009	0.305 ±0.007	0.305 ±0.007

* indicates significant change with $\alpha = 0.05$

4.3 Results

Lateral leg angles θ_R and θ_L in the Fig. 4.2(a) were computed for all eight subjects. The symmetry index of these angles over the group showed significant change with $F(4,28) = 19.287$ and $p < 0.001$. Sphericity was tested in prior as $\chi^2(9) = 9.077$, $p = 0.452$ and pairwise comparison reported significance in BL-T3, BL-T4, T3-PT3, and T4-PT3 pairs. Figure 4.2(c) presents the lateral GRF of subjects. Repeated ANOVA showed significant change of $F(4,20) = 110.761$ and $p < 0.001$ with Sphericity assumed ($\chi^2(9) = 6.092$ and $p = 0.764$).

Gait parameters such as stance phase, step width, and step length were also evaluated to provide a better understanding about the change in the gait pattern, when the intervention was applied. These gait parameters are tabulated in Table 4.1 for each leg as well as the symmetry index. The stance phase became asymmetric during training, where $F(4,28) = 5.241$ and $p = 0.003$ with sphericity test result of $\chi^2(9) = 13.939$, $p = 0.141$. A Significant change was found in step width for both legs. For the step width after right heel strike ($F(1.608, 11.259) = 8.02$, $p = 0.009$), pairwise t-test reported BL-T1 ($p = 0.0002$), BL-T2 ($p = 0.009$), and BL-T3 ($p = 0.022$) as significant pairs. Significant pairs for the step width after left heel strike are reported as BL-T1 ($p = 0.0001$), BL-T2 ($p = 0.009$),and BL-T3 ($p = 0.011$) with $F(1.602, 11.318) = 8.945$, $p = 0.006$. The symmetry index did not show significant difference for step width. Step width was observed to increase presumably because the subjects were likely to step wider in order to increase the base of support [41], to withstand the unnatural asymmetric gait pattern. Step length showed decreasing trend for both legs but there was no significant pairs in the symmetry index for each leg.

Along with gait parameters, Figure 4.3(a-b) present the EMG signals of the gastrocnemius and soleus muscles that play an important role in supporting human gait [48]. From the experiment design, the symmetry index is expected to increase which means the subject is bearing more weight on the right foot over the left foot. First, the significant increase in the asymmetric EMG signal of the gastrocnemius is observed during the training as

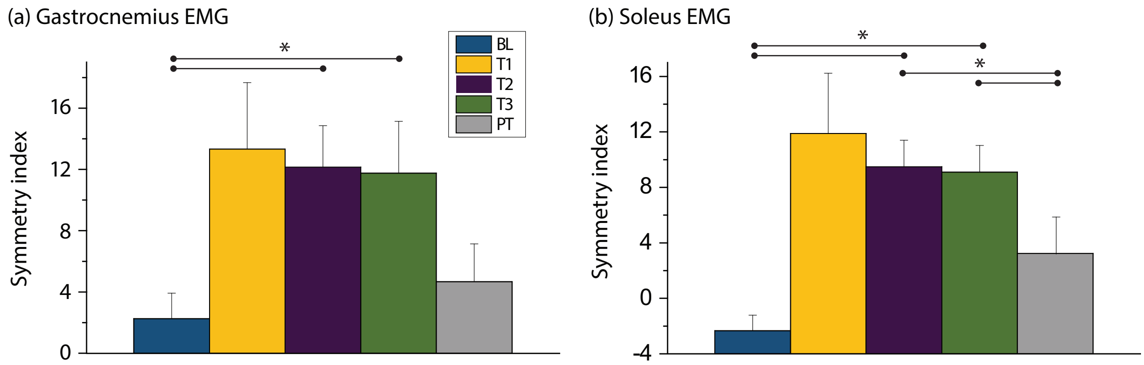


Fig. 4.3 (a) Gastrocnemius and (b) soleus EMG signals demonstrate the significant increases in the asymmetry of weight bearing that is estimated by the symmetry index.

in Fig. 4.3(a): $F(4,28) = 6.215$ and $p = 0.001$. Ahead of repeated ANOVA, sphericity is tested with $\chi^2(9) = 1.557$ and $p = 0.089$. BL-T2 and BL-T3 are reported as significant pairs with Bonferroni-Holmn correction. In Fig. 4.3(b), the soleus muscles are also activated asymmetrically during the training session, where $F(4,28) = 6.862$ and $p = 0.001$. Similarly with gastrocnemius, sphericity was not violated for the soleus either with $\chi^2(9) = 13.882$ and $p = 0.143$. Pairwise t-test reported significant pairs for BL-T2, BL-T3, T2-PT and T3-PT. The integration of ground reaction force is another important measurement because it can estimate weight bearing directly for each foot. Figure 4.2(b) shows that there is a significant increase in the asymmetry of vertical GRF $F(4,28) = 5.901$ and $p = 0.001$. Again, for the vertical GRF, sphericity was not violated with $\chi^2(9) = 14.35$ and $p = 0.126$. The significant pair of GRF was BL-T1.

4.4 Discussion

The free body diagram in Fig. 4.2(a) illustrates the forces acting on the subject during normal walking and TPAD intervention. The TPAD force applied at the pelvic center was mainly to the right side because of the nature of the target trajectory shown in Fig. 4.1(a). While the subjects resisted against being pulled to the right, the lateral GRF of right leg $F_{R,l}$ was increased. This increased lateral ground reaction force on the right leg also increased clockwise moment (from back view) in the coronal plane respect to the pelvic center. To prevent falling to the right side, subjects chose two different strategies. They increased the vertical GRF to resist the moment created by the lateral TPAD force. Another strategy they took was to increase the moment arm of the right leg. The right leg made a wider step than left leg to increase the counter clockwise moment to keep the stability while walking. This observation is demonstrated in Fig. 4.2(c) and Table 4.1. Presumably, due to this same reason, the stance time asymmetrically increased on the right step as presented in Table 4.1. Same observation was also reported in the previous study, where the researcher manually provided a lateral perturbation that impeded the progression of the CoM to the side of the stance leg [89]. The previous study claimed that subjects made a wider step of the swing leg, when the progression of pelvis on the stance leg was prohibited.

The relative increase in the vertical GRF on the right foot causes the right leg naturally to put more effort to bear more weight on the right leg. This claim is supported by the symmetry index of vertical GRF in Fig. 4.2(b) and muscle activation data that bears the weight as seen in Fig. 4.3. Similar to the previous studies, the gastrocnemius and soleus were used as an the indicator to evaluate the weight bearing of diplegic cerebral palsy children [121].

Despite the present demonstration of the changes in the weight loading, further discussion may still be required to translate this novel training to patient groups. Earlier literature has argued that the training effects on healthy subjects and patients are not always same because the biomechanical characteristics and capabilities of patients are different from healthy subjects [25]. The future study with patients is required to modify the control variables in order to make the training efficient. Parameters for the force field profile such as the maximum force value (K_n) or damping constant (K_d) were chosen while a healthy subject was walking on the treadmill. As patients might have damaged stability mechanism and different walking characteristics, an automated module to find optimal parameters should be further investigated. In addition, the training effect of this study on healthy subjects was observed to last for around seven minutes after the device was removed as seen in Fig. 4.3. Future study should be designed to warrant investigation of the training effect for longer observation after the training.

4.5 Conclusion

The present study with healthy subjects provides the feasibility of regulating symmetry of weight bearing by applying a force based on the asymmetric target trajectory of the pelvis. The asymmetric force field is generated by the closed-loop-controller that is implemented in a cable actuated pelvic device. We expect that the present results with eight healthy subjects can be further employed to promote symmetric weight bearing in the gait of hemiplegic patients.

Chapter 5

Concluding remarks

5.1 Contributions of the current work

The scientific contribution of this dissertation is to develop various gait rehabilitation training methods using TPAD. Chapter 2 presented a novel control method to guide and correct pelvic movement. With this new control method, subjects were able to learn a new pelvic trajectory and retain the training effect after tethers were removed. Even though there was no direct intervention on the lower extremities, the movements of the lower limbs also changed in order to be compatible with the target pelvic trajectory. In addition, we found from the results of Experiment III that a training paradigm using both visual and force feedback simultaneously could enhance the training effect. Instead of providing intervention at the lower extremity such as using an exoskeleton type device, TPAD and its controllers can directly intervene at the center of mass to improve impaired gait of patient groups and provide better mobility which is crucial to their quality of life.

Chapter 3 showed a new training paradigm using the TPAD to strengthen lower limbs for correcting the posture of children with crouch gait. In previous studies, robotic devices were targeted to correct the displacement of end-effector (ankle) or joint angles, but this new resistive training paradigm is designed from in-depth understanding of pathology of the crouch gait. The downward pelvic pull was designed to increase the activation of the soleus muscle and delay the early activation of the gastrocnemius muscle. This change of muscle coordination enabled straighter lower limbs during the middle of stance phase and improved walking. The treadmill training with added downward force translated to over ground and we were able to observe these children walk faster and straighter post training. The success of this resistive training with extra downward force underlines the importance of voluntary effort to move their limb and strengthen muscles. We believe that the presented training paradigm suggests a new vision of robotic rehabilitation which can be extended to other pathological gait or impairments.

The last study in this dissertation showed the feasibility of modifying the weight bearing by providing a guidance force in response to pelvic movement. The force was generated by a force field controller with separate right and left modules implemented in TPAD. Eight healthy subjects were trained to walk asymmetrically, similar to stroke patients. As the evidence of modified weight bearing, EMG data on muscles responsible for weight bearing and the ground reaction force from force plates presented significant asymmetric changes during the training. We believe the underlying idea behind this study will help to promote symmetric weight bearing of patients with hemiparesis to prevent the excessive leaning on the non-paretic limb as well as transfer more weight to the paretic limb.

5.2 Suggestions for the future work

The broad goal of the present work in Chapter 2 is to guide and train patients with abnormal pelvic movement. The proposed controller can be used to develop protocols for patients with abnormal pelvic movement. The ability of the system to control the applied force and provide visual feedback may help correct the patients' abnormal pelvic movement to improve their unstable gait. For example, Experiment II can be used to reduce excessive or asymmetric lateral pelvic movement that is often observed in the stroke patients and in the elderly. This controller can train both groups with repeated robotic training to keep their center of mass within the base of support. This alteration of the pelvic movement may reduce falls while walking and enable independent ambulation.

Children with cerebral palsy can also benefit from this new robotic training. Children with cerebral palsy are often observed with excessive pelvic movement compared to typically developing children. The ability to control the pelvic motion will help them in reducing the muscle fatigue and postural imbalance. As introduced before, the target pelvic trajectory of Experiment III is motivated from the comparison study of the pelvic movement between the children with cerebral palsy and typically developing children. The target pelvic trajectory similar to Experiment III can be used to suppress the redundant pelvic movement of cerebral palsy children. In the future, it should be further investigated if subjects will retain the training effect with larger magnitude of force and increased training dosage.

As a next step of this study, patient groups with abnormal pelvic trajectory will be recruited and trained with this novel control method. During the experiment, dynamic balance, lower limb muscle activation, and metabolic energy consumption will be also measured. This pelvic control strategy can be used to control the center of mass during gait rehabilitation or combined with other leg robotic devices for severe patients.

This resistive strength training reported in Chapter 3 is simple and shows feasibility to enhance the quality of walking for children with crouch gait. Because there are no well-established physical therapies or strengthening methods for children with crouch gait, our training method may be a promising intervention for these children. This new training could be used solely as a strength training device in the clinic, or combined with conventional treatment, such as Botox or ethanol injection, to reinforce both methodologies.

There are many strength training devices, but most of them are limited in stationary postures or for specific joint motions, such as leg presses or rubber bands. The key point of the training in Chapter 3 is that the subject can improve their strength during walking by resisting the downward force applied by TPAD. Patient groups other than cerebral palsy with different pathologies who have weak lower limbs can be other good candidates for this study. Patients with abnormal gait tend to walk less after the onset of the disease due to issues of balance, pain, or fatigue. If they walk less, their muscle strength and coordination further reduces. In addition, loss of muscle mass and strength are natural consequences of the aging process, which is known to improve by exercise. We envision that this simple but novel training method can be transferred to a home device installed around a treadmill for different patients and age groups.

References

[1] Af Klint, R., Mazzaro, N., Nielsen, J. B., Sinkjaer, T., and Grey, M. J. (2010). Load rather than length sensitive feedback contributes to soleus muscle activity during human treadmill walking. *J. Neurophysiol.*, 103(5):2747–2756.

[2] Agrawal, S., Kang, J., Chen, X., Kim, M., Lee, Y., Kong, S., and Park, G. (2013). Case studies of a robot enhanced walker for training of children with cerebral palsy. In *IEEE/RSJ International Conference on Intelligent Robots and Systems (IROS)*, pages 4243–4248.

[3] Agrawal, S. K., Kang, J., Chen, X., Kim, M. J., Lee, Y., Kong, S. W., Cho, H., and Park, G.-J. (2016). Robot-enhanced mobility training of children with cerebral palsy: Short-term and long-term pilot studies. *IEEE Syst. J.*, 10(3):1098–1106.

[4] Agrawal, S. K., Vashista, V., Jiyeon, K., and Jin, X. (2015). Human movement research, therapeutic, and diagnostic devices, methods, and systems. US Patent App. 15/304,147.

[5] Ajemian, S., Thon, D., Clare, P., Kaul, L., Zernicke, R. F., and Loitz-Ramage, B. (2004). Cane-assisted gait biomechanics and electromyography after total hip arthroplasty. *Arch. Phys. Med. Rehabil.*, 85(12):1966–1971.

[6] Allemand, Y., Stauffer, Y., Clavel, R., and Brodard, R. (2009). Design of a new lower extremity orthosis for overground gait training with the walktrainer. In *International Conference on Rehabilitation Robotics(ICORR)*, pages 550–555.

[7] Anttila, H., Autti-Rämö, I., Suoranta, J., Mäkelä, M., and Malmivaara, A. (2008). Effectiveness of physical therapy interventions for children with cerebral palsy: a systematic review. *BMC Pediatr.*, 8(1):1.

[8] Aoyagi, D., Ichinose, W. E., Harkema, S. J., Reinkensmeyer, D. J., and Bobrow, J. E. (2007). A robot and control algorithm that can synchronously assist in naturalistic motion during body-weight-supported gait training following neurologic injury. *IEEE Trans. Neural Syst. Rehabil. Eng.*, 15(3):387–400.

[9] Awad, L. N., Bae, J., O'donnell, K., De Rossi, S. M., Hendron, K., Sloot, L. H., Kudzia, P., Allen, S., Holt, K. G., Ellis, T. D., and Walsh, C. J. (2017). A soft robotic exosuit improves walking in patients after stroke. *Sci. Transl. Med.*, 9(400):eaai9084.

[10] Banala, S. K., Kim, S. H., Agrawal, S. K., and Scholz, J. P. (2009). Robot assisted gait training with active leg exoskeleton (alex). *IEEE Trans. Neural Syst. Rehabil. Eng.*, 17(1):2–8.

[11] Bastiaanse, C., Duysens, J., and Dietz, V. (2000). Modulation of cutaneous reflexes by load receptor input during human walking. *Exp. Brain Res.*, 135(2):189–198.

[12] Begnoche, D. M. and Pitetti, K. H. (2007). Effects of traditional treatment and partial body weight treadmill training on the motor skills of children with spastic cerebral palsy: a pilot study. *Pediatr. Phys. Ther.*, 19(1):11–19.

[13] Bennett, B. C., Abel, M. F., Wolovick, A., Franklin, T., Allaire, P. E., and Kerrigan, D. C. (2005). Center of mass movement and energy transfer during walking in children with cerebral palsy. *Arch. Phys. Med. Rehabil.*, 86(11):2189–2194.

[14] Binder, H. and Eng, G. D. (1989). Rehabilitation management of children with spastic diplegic cerebral palsy. *Arch. Phys. Med. Rehabil.*, 70(6):482–9.

[15] Blundell, S., Shepherd, R., Dean, C., Adams, R., and Cahill, B. (2003). Functional strength training in cerebral palsy: a pilot study of a group circuit training class for children aged 4–8 years. *Clin. Rehabil.*, 17(1):48–57.

[16] Cai, L. L., Fong, A. J., Otoshi, C. K., Liang, Y., Burdick, J. W., Roy, R. R., and Edgerton, V. R. (2006). Implications of assist-as-needed robotic step training after a complete spinal cord injury on intrinsic strategies of motor learning. *Journal of Neuroscience*, 26(41):10564–10568.

[17] Cappozzo, A. (1981). Analysis of the linear displacement of the head and trunk during walking at different speeds. *J. Biomech.*, 14(6):411–425.

[18] Chen, G., Patten, C., Kothari, D. H., and Zajac, F. E. (2005). Gait deviations associated with post-stroke hemiparesis: improvement during treadmill walking using weight support, speed, support stiffness, and handrail hold. *Gait Posture*, 22(1):57–62.

[19] Cherng, R.-J., Liu, C.-F., Lau, T.-W., and Hong, R.-B. (2007). Effect of treadmill training with body weight support on gait and gross motor function in children with spastic cerebral palsy. *Am. J. Phys. Med. Rehabil.*, 86(7):548–555.

[20] Chow, D. H., Kwok, M. L., Au-Yang, A. C., Holmes, A. D., Cheng, J. C., Yao, F. Y., and Wong, M. (2005). The effect of backpack load on the gait of normal adolescent girls. *Ergonomics*, 48(6):642–656.

[21] Chrysagis, N., Skordilis, E. K., Stavrou, N., Grammatopoulou, E., and Koutsouki, D. (2012). The effect of treadmill training on gross motor function and walking speed in ambulatory adolescents with cerebral palsy: a randomized controlled trial. *Am. J. Phys. Med. Rehabil.*, 91(9):747–760.

[22] Damiano, D. L., Arnold, A. S., Steele, K. M., and Delp, S. L. (2010). Can strength training predictably improve gait kinematics? a pilot study on the effects of hip and

knee extensor strengthening on lower-extremity alignment in cerebral palsy. *Phys. Ther.*, 90(2):269–279.

[23] Damiano, D. L., Kelly, L. E., and Vaughn, C. L. (1995a). Effects of quadriceps femoris muscle strengthening on crouch gait in children with spastic diplegia. *Phys. Ther.*, 75(8):658–667.

[24] Damiano, D. L., Vaughan, C. L., and Abel, M. E. (1995b). Muscle response to heavy resistance exercise in children with spastic cerebral palsy. *Dev. Med. Child Neurol.*, 37(8):731–739.

[25] Dean, J. C. and Kautz, S. A. (2015). Foot placement control and gait instability among people with stroke. *J. Rehabil. Res. Dev.*, 52(5):577.

[26] Dietz, V. and Berger, W. (1995). Cerebral palsy and muscle transformation. *Dev. Med. Child. Neurol.*, 37(2):180–184.

[27] Dodd, K. J. and Foley, S. (2007). Partial body-weight-supported treadmill training can improve walking in children with cerebral palsy: a clinical controlled trial. *Dev. Med. Child Neurol.*, 49(2):101–105.

[28] Dodd, K. J. and Morris, M. E. (2003). Lateral pelvic displacement during gait: abnormalities after stroke and changes during the first month of rehabilitation. *Arch. Phys. Med. Rehabil.*, 84(8):1200–1205.

[29] Donelan, J. M., Kram, R., and Kuo, A. D. (2001). Mechanical and metabolic determinants of the preferred step width in human walking. *Proc. R. Soc. B*, 268:1985–1992.

[30] Donelan, J. M., Shipman, D. W., Kram, R., and Kuo, A. D. (2004). Mechanical and metabolic requirements for active lateral stabilization in human walking. *J. Biomech.*, 37(6):827–835.

[31] Duschau-Wicke, A., Caprez, A., and Riener, R. (2010). Patient-cooperative control increases active participation of individuals with sci during robot-aided gait training. *J. Neuroeng. Rehabil.*, 7(43):1–13.

[32] Eek, M. N., Tranberg, R., Zügner, R., Alkema, K., and Beckung, E. (2008). Muscle strength training to improve gait function in children with cerebral palsy. *Dev. Med. Child Neurol.*, 50(10):759–764.

[33] Emken, J. L., Benitez, R., and Reinkensmeyer, D. J. (2007). Human-robot cooperative movement training: learning a novel sensory motor transformation during walking with robotic assistance-as-needed. *Journal of NeuroEngineering and Rehabilitation*, 4(1):8.

[34] Gage, J. R. (1991). *Gait analysis in cerebral palsy*. Mac Keith Press.

[35] Go, A. S., Mozaffarian, D., Roger, V. L., Benjamin, E. J., Berry, J. D., Borden, W. B., Bravata, D. M., Dai, S., Ford, E. S., Fox, C. S., et al. (2013). Executive summary: heart disease and stroke statistics—2013 update: a report from the american heart association. *Circulation*, 127(1):143–152.

[36] Harkema, S. J., Hurley, S. L., Patel, U. K., Requejo, P. S., Dobkin, B. H., and Edgerton, V. R. (1997). Human lumbosacral spinal cord interprets loading during stepping. *J. Neurophysiol.*, 77(2):797–811.

[37] Hesse, S., Jahnke, M., Schreiner, C., and Mauritz, K. (1993). Gait symmetry and functional walking performance in hemiparetic patients prior to and after a 4-week rehabilitation programme. *Gait Posture*, 1(3):166–171.

[38] Hesse, S. A., Jahnke, M. T., Bertelt, C. M., Schreiner, C., Lücke, D., and Mauritz, K.-H. (1994). Gait outcome in ambulatory hemiparetic patients after a 4-week comprehensive rehabilitation program and prognostic factors. *Stroke*, 25(10):1999–2004.

[39] Hodapp, M., Vry, J., Mall, V., and Faist, M. (2009). Changes in soleus h-reflex modulation after treadmill training in children with cerebral palsy. *Brain*, 132(1):37–44.

[40] Hof, A., Elzinga, H., Grimmius, W., and Halbertsma, J. (2002). Speed dependence of averaged EMG profiles in walking. *Gait & posture*, 16(1):78–86.

[41] Hof, A., Vermerris, S., and Gjaltema, W. (2010). Balance responses to lateral perturbations in human treadmill walking. *J. Exp. Biol.*, 213(15):2655–2664.

[42] Hsu, C.-J., Kim, J., Tang, R., Roth, E. J., Rymer, W. Z., and Wu, M. (2017). Applying a pelvic corrective force induces forced use of the paretic leg and improves paretic leg emg activities of individuals post-stroke during treadmill walking. *Clin. Neurophysiol.*, 128(10):1915–1922.

[43] Hsue, B.-J., Miller, F., and Su, F.-C. (2009). The dynamic balance of the children with cerebral palsy and typical developing during gait. part 1: Spatial relationship between com and cop trajectories. *Gait Posture*, 29(3):465–470.

[44] Husemann, B., Müller, F., Krewer, C., Heller, S., and Koenig, E. (2007). Effects of locomotion training with assistance of a robot-driven gait orthosis in hemiparetic patients after stroke. *Stroke*, 38(2):349–354.

[45] Jarus, T. and Gutman, T. (2001). Effects of cognitive processes and task complexity on acquisition, retention, and transfer of motor skills. *Can. J. Occup. Ther.*, 68(5):280–289.

[46] Kang, J., Kim, K., Kim, S. J., and Oh, S. R. (2012). On the comfortableness of muscle power assistive robotic system. In *IEEE RAS & EMBS International Conference on Biomedical Robotics and Biomechatronics (BioRob)*, pages 1759–1764.

[47] Kang, J., Logan, S., Galloway, J. C., and Agrawal, S. K. (2014). A chase-game to teach children on a robot to follow moving objects. In *Proc. IEEE Int. Conf. Robot. Autom. (ICRA)*, pages 234–239.

[48] Kang, J., Martelli, D., Vashista, V., Martinez-Hernandez, I., Kim, H., and Agrawal, S. K. (2017a). Robot-driven downward pelvic pull to improve crouch gait in children with cerebral palsy. *Sci. Robot.*, 2(8):eaan2634.

[49] Kang, J., Vashista, V., and Agrawal, S. (2017b). On the adaptation of pelvic motion by applying 3-dimensional guidance forces using TPAD. *IEEE Trans. Neural Syst. Rehabil. Eng.*, 25(9):1558–1567.

[50] Kang, J., Vashista, V., and Agrawal, S. K. (2015). A novel assist-as-needed control method to guide pelvic trajectory for gait rehabilitation. In *IEEE International Conference on Rehabilitation Robotics (ICORR)*, pages 630–635.

[51] Kedem, P. and Scher, D. M. (2016). Evaluation and management of crouch gait. *Curr. Opin. Pediatr.*, 28(1):55–59.

[52] Kembhavi, G., Darrah, J., Magill-Evans, J., and Loomis, J. (2002). Using the berg balance scale to distinguish balance abilities in children with cerebral palsy. *Pediatr. Phys. Ther.*, 14(2):92–99.

[53] Kim, C. M. and Eng, J. J. (2003). Symmetry in vertical ground reaction force is accompanied by symmetry in temporal but not distance variables of gait in persons with stroke. *Gait Posture*, 18(1):23–28.

[54] Kim, S. H., Banala, S. K., Brackbill, E. A., Agrawal, S. K., Krishnamoorthy, V., and Scholz, J. P. (2010). Robot-assisted modifications of gait in healthy individuals. *Exp. Brain Res.*, 202(4):809–824.

[55] Kirker, S., Simpson, D., Jenner, J., and Wing, A. (2000). Stepping before standing: hip muscle function in stepping and standing balance after stroke. *J. Neurol. Neurosurg. Psychiatry*, 68(4):458–464.

[56] Knox, V. and Evans, A. L. (2002). Evaluation of the functional effects of a course of bobath therapy in children with cerebral palsy: a preliminary study. *Dev. Med. Child Neurol.*, 44(7):447–460.

[57] Kuo, A. D. (1999). Stabilization of lateral motion in passive dynamic walking. *Int. J. Robot. Res.*, 18(9):917–930.

[58] Lam, T., Anderschitz, M., and Dietz, V. (2006). Contribution of feedback and feedforward strategies to locomotor adaptations. *J. Neurophysiol.*, 95(2):766–773.

[59] Lam, T., Luttmann, K., Houldin, A., and Chan, C. (2009). Treadmill-based locomotor training with leg weights to enhance functional ambulation in people with chronic stroke: a pilot study. *J. Neurol. Phys. Ther.*, 33(3):129–135.

[60] Leunkeu, A. N., Lelard, T., Shephard, R. J., Doutrellot, P.-L., and Ahmaidi, S. (2014). Reproducibility of gait cycle and plantar pressure distribution in children with spastic hemiplegic cerebral palsy. *NeuroRehabilitation*, 35(3):597–606.

[61] Li, J. X., Hong, Y., and Robinson, P. D. (2003). The effect of load carriage on movement kinematics and respiratory parameters in children during walking. *European Journal of Applied Physiology*, 90(1-2):35–43.

[62] Lorentzen, J., Kirk, H., Fernandez-Lago, H., Frisk, R., Scharff Nielsen, N., Jorsal, M., and Nielsen, J. B. (2017). Treadmill training with an incline reduces ankle joint stiffness and improves active range of movement during gait in adults with cerebral palsy. *Disability and rehabilitation*, 39(10):987–993.

[63] Luu, T. P., Low, K. H., Qu, X., Lim, H. B., and Hoon, K. H. (2014). Hardware development and locomotion control strategy for an over-ground gait trainer: Nature-gaits. *IEEE J. Transl. Eng. Health Med.*, 2:1–9.

[64] MacKinnon, C. D. and Winter, D. A. (1993). Control of whole body balance in the frontal plane during human walking. *J. Biomech.*, 26(6):633–644.

[65] Magill, R. A. and Anderson, D. I. (2007). *Motor learning and control: Concepts and applications*, volume 11. McGraw-Hill New York.

[66] Massaad, F., Lejeune, T. M., and Detrembleur, C. (2007). The up and down bobbing of human walking: a compromise between muscle work and efficiency. *J. Physiol*, 582(2):789–799.

[67] Mauritz, K.-H. (2002). Gait training in hemiplegia. *Eur. J. Neurol.*, 9(s1):23–29.

[68] McGowan, C., Kram, R., and Neptune, R. (2009). Modulation of leg muscle function in response to altered demand for body support and forward propulsion during walking. *J. Biomech.*, 42(7):850–856.

[69] McNeill Alexander, R. (2002). Energetics and optimization of human walking and running: the 2000 raymond pearl memorial lecture. *Am. J. Hum. Biol*, 14(5):641–648.

[70] Meuleman, J. H., Van Asseldonk, E. H., and Van der Kooij, H. (2013). The effect of directional inertias added to pelvis and ankle on gait. *J. Neuroeng. Rehabil.*, 10(40):1–12.

[71] Michaud, T. C. (2011). *Human locomotion: the conservative management of gait-related disorders*. Newton Biomechanics.

[72] Ming, A. and Higuchi, T. (1994). Study on multiple degree-of-freedom positioning mechanism using wires. 1: Concept, design and control. *Int. J. Jpn. Soc. Precis. Eng.*, 28(2):131–138.

[73] Morton, J. F., Brownlee, M., and McFadyen, A. K. (2005). The effects of progressive resistance training for children with cerebral palsy. *Clin. Rehabil.*, 19(3):283–289.

[74] Mudge, S., Rochester, L., and Recordon, A. (2003). The effect of treadmill training on gait, balance and trunk control in a hemiplegic subject: a single system design. *Disabil. Rehabil.*, 25(17):1000–1007.

[75] Mun, K.-R., Lim, S. B., Guo, Z., and Yu, H. (2017). Biomechanical effects of body weight support with a novel robotic walker for over-ground gait rehabilitation. *Med. Biol. Eng. Comput.*, 55(2):315–326.

[76] Munawar, H., Yalcin, M., and Patoglu, V. (2016). Redundant kinematics and workspace centering control of assiston-gait overground gait and balance trainer. In *Proc. IEEE Int. Conf. Robot. Autom. (ICRA)*, pages 3704–3710.

[77] Mustafa, S. K. and Agrawal, S. K. (2012). On the force-closure analysis of n-dof cable-driven open chains based on reciprocal screw theory. *IEEE Trans. Robot.*, 28(1):22–31.

[78] Neumann, D. A. (2011). *Kinesiology of the musculoskeletal system: foundations for rehabilitation. St Louis, MO: Mosby*. Elsevier.

[79] Nicolini-Panisson, R. D. and Donadio, M. V. F. (2013). Timed" up & go" test in children and adolescents. *Rev. Paul. Pediatr.*, 31(3):377–383.

[80] Norlin, R. and Odenrick, P. (1986). Development of gait in spastic children with cerebral palsy. *J. Pediatr. Orthop.*, 6(6):674–680.

[81] Olenšek, A., Zadravec, M., and Matjačić, Z. (2016). A novel robot for imposing perturbations during overground walking: mechanism, control and normative stepping responses. *J. Neuroeng. Rehabil.*, 13(1):55.

[82] Pai, Y.-C. and Patton, J. (1997). Center of mass velocity-position predictions for balance control. *J. Biomech.*, 30(4):347–354.

[83] Patterson, K. K., Mansfield, A., Biasin, L., Brunton, K., Inness, E. L., and McIlroy, W. E. (2015). Longitudinal changes in poststroke spatiotemporal gait asymmetry over inpatient rehabilitation. *Neurorehabil. Neural Repair*, 29(2):153–162.

[84] Patterson, K. K., Parafianowicz, I., Danells, C. J., Closson, V., Verrier, M. C., Staines, W. R., Black, S. E., and McIlroy, W. E. (2008). Gait asymmetry in community-ambulating stroke survivors. *Arch. Phys. Med. Rehabil.*, 89(2):304–310.

[85] Perry, J., Davids, J. R., et al. (1992). Gait analysis: normal and pathological function. *J. Pediatr. Orthop.*, 12(6):815.

[86] Peshkin, M., Brown, D. A., Santos-Munné, J. J., Makhlin, A., Lewis, E., Colgate, J. E., Patton, J., and Schwandt, D. (2005). KineAssist: A robotic overground gait and balance

training device. In *International Conference on Rehabilitation Robotics(ICORR)*, pages 241–246.

[87] Pietrusinski, M., Cajigas, I., Severini, G., Bonato, P., and Mavroidis, C. (2014a). Robotic Gait Rehabilitation Trainer. *IEEE/ASME Trans. Mechatron.*, 19(2):490–499.

[88] Pietrusinski, M., Cajigas, I., Severini, G., Bonato, P., and Mavroidis, C. (2014b). Robotic gait rehabilitation trainer. *IEEE/ASME Trans. Mechatronics*, 19(2):490–499.

[89] Rankin, B. L., Buffo, S. K., and Dean, J. C. (2014). A neuromechanical strategy for mediolateral foot placement in walking humans. *J. Neurophysiol.*, 112(2):374–383.

[90] Regnaux, J., Pradon, D., Roche, N., Robertson, J., Bussel, B., and Dobkin, B. (2008). Effects of loading the unaffected limb for one session of locomotor training on laboratory measures of gait in stroke. *Clin. Biomech.*, 23(6):762–768.

[91] Robinovitch, S. N., Feldman, F., Yang, Y., Schonnop, R., Leung, P. M., Sarraf, T., Sims-Gould, J., and Loughin, M. (2013). Video capture of the circumstances of falls in elderly people residing in long-term care: an observational study. *Lancet*, 381(9860):47–54.

[92] Rodda, J., Graham, H., Carson, L., Galea, M., and Wolfe, R. (2004). Sagittal gait patterns in spastic diplegia. *Bone Joint J.*, 86(2):251–258.

[93] Rose, J. (2009). Selective motor control in spastic cerebral palsy. *Dev. Med. Child Neurol.*, 51(8):578–579.

[94] Saunders, J. B., Inman, V. T., Eberhart, H. D., et al. (1953). The major determinants in normal and pathological gait. *J. Bone Joint Surg.*, 35(3):543–558.

[95] Schindl, M. R., Forstner, C., Kern, H., and Hesse, S. (2000). Treadmill training with partial body weight support in nonambulatory patients with cerebral palsy. *Arch. Phys. Med. Rehabil.*, 81(3):301–306.

[96] Scholtes, V. A., Becher, J. G., Janssen-Potten, Y. J., Dekkers, H., Smallenbroek, L., and Dallmeijer, A. J. (2012). Effectiveness of functional progressive resistance exercise training on walking ability in children with cerebral palsy: a randomized controlled trial. *Res. Dev. Disabil.*, 33(1):181–188.

[97] Scholtes, V. A., Dallmeijer, A. J., Rameckers, E. A., Verschuren, O., Tempelaars, E., Hensen, M., and Becher, J. G. (2008). Lower limb strength training in children with cerebral palsy–a randomized controlled trial protocol for functional strength training based on progressive resistance exercise principles. *BMC Pediatr.*, 8(1):41.

[98] Scianni, A., Butler, J. M., Ada, L., and Teixeira-Salmela, L. F. (2009). Muscle strengthening is not effective in children and adolescents with cerebral palsy: a systematic review. *Aust. J. Physiother.*, 55(2):81–87.

[99] Sheikh, M., Azarpazhooh, M. R., and Hosseini, H. A. (2016). Randomized comparison trial of gait training with and without compelled weight-shift therapy in individuals with chronic stroke. *Clin. Rehabil.*, 30(11):1088–1096.

[100] Shirota, C., Van Asseldonk, E., Vallery, H., Buurke, J. H., Veneman, J. F., Barralon, P., Maggioni, S., and Matjačić, Z. (2017). Robot-supported assessment of balance in standing and walking. *J. Neuroeng. Rehabil.*, 14(1):80.

[101] Srivastava, S., Kao, P. C., Reisman, D. S., Scholz, J. P., Agrawal, S. K., and Higginson, J. S. (2016). Robotic assist-as-needed as an alternative to therapist-assisted gait rehabilitation. *International journal of physical medicine & rehabilitation*, 4(5).

[102] Steele, K., Damiano, D., Eek, M., Unger, M., and Delp, S. (2012). Characteristics associated with improved knee extension after strength training for individuals with cerebral palsy and crouch gait. *J. Pediatr. Rehabil. Med.*, 5(2):99–106.

[103] Steele, K. M., Seth, A., Hicks, J. L., Schwartz, M. H., and Delp, S. L. (2013). Muscle contributions to vertical and fore-aft accelerations are altered in subjects with crouch gait. *Gait posture*, 38(1):86–91.

[104] Steele, K. M., Seth, A., Hicks, J. L., Schwartz, M. S., and Delp, S. L. (2010). Muscle contributions to support and progression during single-limb stance in crouch gait. *J. Biomech.*, 43(11):2099–2105.

[105] Sutherland, D. H. and Davids, J. R. (1993). Common gait abnormalities of the knee in cerebral palsy. *Clin. Orthop. Relat. Res.*, 288:139–147.

[106] Thompson, P., Beath, T., Bell, J., Jacobson, G., Phair, T., Salbach, N. M., and Wright, F. (2008). Test–retest reliability of the 10-metre fast walk test and 6-minute walk test in ambulatory school-aged children with cerebral palsy. *Dev. Med. Child Neurol.*, 50(5):370–376.

[107] Unger, M., Faure, M., and Frieg, A. (2006). Strength training in adolescent learners with cerebral palsy: a randomized controlled trial. *Clin. Rehabil.*, 20(6):469–477.

[108] van Hedel, H. J., Meyer-Heim, A., and Rüsch-Bohtz, C. (2016). Robot-assisted gait training might be beneficial for more severely affected children with cerebral palsy. *Dev. Neurorehabil.*, 19(6):410–415.

[109] Van Loo, M., Moseley, A., Bosman, J., De Bie, R., and Hassett, L. (2004). Test–re-test reliability of walking speed, step length and step width measurement after traumatic brain injury: a pilot study. *Brain Inj.*, 18(10):1041–1048.

[110] Vashista, V. (2015). *A Cable-Driven Pelvic Robot: Human Gait Adaptation and Rehabilitation Studies*. Columbia University.

[111] Vashista, V., Jin, X., and Agrawal, S. K. (2014). Active tethered pelvic assist device (A-TPAD) to study force adaptation in human walking. In *Proc. IEEE Int. Conf. Robot. Autom. (ICRA)*, pages 718–723.

[112] Vashista, V., Martelli, D., and Agrawal, S. (2016a). Locomotor adaptation to an asymmetric force on the human pelvis directed along the right leg. *IEEE Trans. Neural Syst. Rehabil. Eng.*, 24(8):872–881.

[113] Vashista, V., Martelli, D., and Agrawal, S. K. (2016b). Locomotor adaptation to an asymmetric force on the human pelvis directed along the right leg. *IEEE Trans. Neural Syst. Rehabil. Eng.*, 24(8):872–881.

[114] Whittle, M. W. (1997). Three-dimensional motion of the center of gravity of the body during walking. *Hum. Mov. Sci.*, 16(2):347–355.

[115] Willerslev-Olsen, M., Lorentzen, J., and Nielsen, J. B. (2014). Gait training reduces ankle joint stiffness and facilitates heel strike in children with cerebral palsy. *NeuroRehabilitation*, 35(4):643–655.

[116] Willerslev-Olsen, M., Petersen, T. H., Farmer, S. F., and Nielsen, J. B. (2015). Gait training facilitates central drive to ankle dorsiflexors in children with cerebral palsy. *Brain*, 138(3):589–603.

[117] Williams, E. N., Carroll, S. G., Reddihough, D. S., Phillips, B. A., and Galea, M. P. (2005). Investigation of the timed 'up & go'test in children. *Dev. Med. Child Neurol.*, 47(8):518–524.

[118] Winter, D. A. (1991). *Biomechanics and motor control of human gait: normal, elderly and pathological*. University of Waterloo Press.

[119] Winter, D. A. (2009). *Biomechanics and motor control of human movement*. John Wiley & Sons.

[120] Winter, S., Autry, A., Boyle, C., and Yeargin-Allsopp, M. (2002). Trends in the prevalence of cerebral palsy in a population-based study. *Pediatrics*, 110(6):1220–1225.

[121] Wu, M., Kim, J., Arora, P., Gaebler-Spira, D. J., and Zhang, Y. (2016). Kinematic and EMG responses to pelvis and leg assistance force during treadmill walking in children with cerebral palsy. *Neural Plast.*, 2016.

[122] Young, P. M. M., Wilken, J. M., and Dingwell, J. B. (2012). Dynamic margins of stability during human walking in destabilizing environments. *J. Biomech.*, 45(6):1053–1059.

[123] Zar, J. H. (1984). *Biostatistical analysis. 2nd.* Prentice Hall USA.

[124] Zeni, J., Richards, J., and Higginson, J. (2008). Two simple methods for determining gait events during treadmill and overground walking using kinematic data. *Gait Posture*, 27(4):710–714.

[125] Zijlstra, W. and Hof, A. L. (1997). Displacement of the pelvis during human walking: experimental data and model predictions. *Gait Posture*, 6(3):249–262.